WHICH PROPRIOCEPTION ASSESSMENT METHOD BEST PREDICTS BALANCE ABILITY IN HEALTHY YOUNG AND ELDERLY POPULATIONS?

By

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A thesis submitted to

The University of Birmingham

For the degree of

DOCTOR OF PHILOSOPHY

School of Sport, Exercise and Rehabilitation Sciences

Collage of Life and Environmental Sciences

University of Birmingham

August 2023

Abstract

Determining the process through which balance and ankle proprioception are linked has been a relevant topic in the last decade due to the direct clinical implications associated with balance deficits. The overall aim of this thesis was to determine which proprioceptive assessment method best predicts balance ability in healthy young and elderly populations. We started our research with a detailed literature review on proprioception (Chapter 1) to assess the gaps of knowledge in the database on proprioception and its link to balance. We aimed to compare different proprioceptive assessment methods to determine which one best predicted balance performance. We therefore started by developing several proprioceptive assessment methods (described in Chapter 2). We were able to develop two additional assessment methods to test in our balance lab beside joint position reproduction (JPR), which had been developed prior to the commencement of our research. Both assessment methods were variations of the threshold to detection of passive motion (TTDPM) method. The first measured detection threshold as the amplitude at which passive ankle motion was detected, and the second measured reaction time of participants the moment they detected ankle motion. Using these techniques, we performed a series of experimental studies. The first experimental study (Chapter 3) determined whether age-related decline in ankle proprioception, as measured through a JPR task, is affected by muscle strength. Our results indicated that while age significantly affected proprioceptive acuity at the ankle, muscle strength had a significant effect on the proprioceptive performance of middle-aged participants only. Our second experimental study (Chapter 4) established whether isometric contraction influenced ankle proprioception.

We found that while there was no difference in ankle proprioceptive acuity between a passive and an isometrically contracted leg, the stability of muscular force output–as measured through torque variability–significantly predicted proprioceptive performance. Our third experimental study (Chapter 5), determined whether ankle proprioception, measured using a reaction-time method, was predictive of postural sway during quiet standing. We found that reaction time significantly predicted standing balance in the anterior-posterior plane when feet were apart. We also found that age significantly correlated with reaction time, with older participants reacting at significantly slower rates than younger participants. In addition, we found a significant interaction between age and vision when we analysed body sway data separately. This indicates that older aged participants relied on vision more than younger participants to maintain their balance during quiet standing. Our findings show that age significantly predicts proprioceptive acuity at the ankle, with older participants consistently performing worse using different proprioceptive assessment methods. They also suggest that more studies are needed on the effect of muscle strength on proprioceptive performance, as our results indicate there is a link between the two.

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List of Abbreviations

Abbreviation	***Definition***
°	Degrees
°/s	Degrees per Second
ADLs	Activities of Daily Living
AE	Absolute Error
AMEDA	Active Movement Extent Discrimination Assessment
ANOVA	Analysis of Variance

AP	Anterior-Posterior
BBS	Berg Balance Scale
BMI	Body Mass Index
BSw	Body Sway
CE	Constant Error
CJPR	Contralateral Joint Position Reproduction
cm	Centimetres
CNS	Central Nervous System
COP	Centre of Pressure
dc	Down-correct
DF	Dorsiflexion
df	Down-false
EC	Eyes closed
ECFA	Eyes closed feet apart
ECFT	Eyes closed feet together
EMG	Electromyography
EO	Eyes Open
EOFA	Eyes Open-Feet Apart
EOFT	Eyes Open-Feet Together
FES	Falls Efficacy Scale
GBS	Guillain-Barré Syndrome
H_0	Null Hypothesis
H_a	Alternative Hypothesis
Hz	Hertz
ICC	Intra-Class Correlation Coefficient
IJPR	Ipsilateral Joint Position Reproduction
JPM	Joint Position Matching
JPR	Joint Position Reproduction
JT	Joint Torque
kg	Kilograms
m	Metre

M	Mean
MABC	Movement Assessment Battery for Children
ML	Medio-Lateral
mm/s	Millimetres per Second
MVC	Maximum Voluntary Contraction
MWT	Metre Walking Test
N	Newtons
n	Sample size
n.d.	No Date
N.m	Newton-metre
N/A	Not Applicable
PD	Parkinson’s Disease
PDQ	Parkinson’s Disease Questionnaire
PE	Position Error
PEV	Position Error Variability
PF	Plantar Flexion
RMS	Root Mean Square
RTm	Reaction Time
RVC	Reference Voluntary Contraction
s	Second
SD	Standard Deviation
SPPB	Short Physical Performance Battery
STS	Sit to Stand
TTDPM	Threshold to Detection of Passive Motion
TTPPM	Threshold to Perception of Passive Motion
uc	Up-correct
uf	Up-false
VE	Variable Error
yrs	Years
Δ^{+}	Step-up
Δ^{-}	Step-down

Acknowledgments

I would like to express my sincere gratitude to my supervisor, Raymond Reynolds, for his invaluable guidance and support throughout my PhD programme. I would also like to thank my mother and father for their unwavering support throughout my time living overseas in the UK. Lastly, to my husband Ahmed and two daughters Aisha and Munirah: I love you and could not have done this without you.

Chapter 1: General Introduction

Abstract

For our general introduction chapter, we discuss the history of proprioception, how the term was coined, and present-day inconsistencies regarding the use of terms related to proprioception. We go on to describe the types of proprioceptive sensation known at present and the diverse ways in which proprioceptive sensation is divided in current literature, all in an effort to provide an accurate definition of proprioception. A description of the various receptors of proprioception is then presented with figures to view anatomical details of each receptor. After which we discuss the available evidence in support of muscle spindles as the main receptor of proprioception. A brief section on the physiological representation of proprioception in the central nervous system is then presented, before we discuss the role of proprioception in motor learning and the acquisition of new skills. We go on to discuss the relation between proprioception and postural control and the clinical relevance of balance impairments. Studies on standing balance and proprioception are then discussed, after which we present the proprioceptive testing methods most prevalent in the literature. Our literature review linking balance and proprioception follows, in which we discuss evidence on the relation between balance and proprioception in both healthy and clinical populations. A table summarizing our literature review is also provided. Finally, we conclude the chapter with our research question "**Which proprioception assessment method best predicts balance ability in healthy young and elderly populations?".** Our general experimental and null hypotheses are also included followed by a brief description of the flow of our research.

1.1 What is Proprioception?

1.1.1 History

In 1826, Charles Bell determined that a "sixth sense", or a "muscular sense" existed from which the ability to determine the position of one's body and limbs originates. According to Bell, 'Between the brain and the muscles there is a circle of nerves; one nerve [ventral roots] conveys the influence from the brain to the muscle, another [dorsal roots] gives the sense of the condition of the muscle to the brain' (Bell, 1826).

In 1880, H. Charlton Bastian, a London neurologist, introduced the term "kinaesthesia" which is made up of two Greek words: kinein (move) and aesthesis (sensation) and refers to the sensation of limb position and movement (Bastian, 1887).

This term, sense of movement, was found to be preferable in the long run to scientists over the term "muscular sense" as it widens the scope of the components within the body needed to accurately describe this ability. Presently, these would include visual, auditory, cutaneous, vestibular and, lastly, muscular components (Smith, 2011).

However, the actual word "proprioception" was originated in 1902 by Sir Charles Sherrington, an English neurophysiologist, from the Latin word for "one's own" (proprius) and perception (Sherrington, 1907). From his perspective, proprioception referred to the sensory information received from numerous neural receptors located within joints, tendons, and muscles, and which allowed a person to be aware of the location of limbs (and body parts in general) at any given time.

1.1.2 The present-day inconsistency of terms regarding proprioception:

Going through the literature related to proprioception, one may become overwhelmed given the variety in current use of terms related to the proprioceptive system, particularly with regard to the terms “proprioception” and “kinaesthesia”. According to some researchers, proprioception refers to joint *position* sense specifically, whereas kinaesthesia refers to the recognition of joint *motion* or movement sense (Warner et al., 1996; Gardner et al., 2000; Swanik et al., 2004). And in other instances, kinaesthesia is referred to as a *component* of proprioception, with proprioception interpreted as the ability to sense both joint position and joint movement (Ergen and Ulkar, 2008; Wingert et al., 2009). Then again, proprioception defined in this way coincides with Bastian’s interpretation of kinaesthesia which, as mentioned above, includes both joint position and movement senses. Add to that the repetitive use of the term “kinaesthetic sensation” in the literature to describe position and movement sense (Riemann and Lephart, 2002; Proske and Gandevia, 2012) making the two terms “proprioception” and “kinaesthesia” arguably interchangeable, and used as synonyms in many instances (Proske and Gandevia, 2012; Han et al., 2016). However, this leaves other proprioceptive sensations in the literature–such as the senses of resistance, effort, heaviness, and vibration–unclassified, suggesting they are unrelated to position and movement sense, which is not the case (Stillman, 2002). As such, use of the terms proprioception and kinaesthesia synonymously may be inappropriate for the most part. Also, for the sake of clarity, it may be best to use proprioception/proprioceptive senses as a general term, and kinaesthesia/kinaesthetic senses only when describing joint and movement senses specifically.

Moreover, it is believed that proprioceptive senses can be appreciated both consciously and unconsciously, however there are also various discrepancies about term hierarchy regarding this in the literature. In a comprehensive review on the sensorimotor system, Riemann and Lephart (2002) stated that “conscious proprioceptive senses”, including kinaesthesia, joint position sense and the sense of resistance or force, are categorised under the umbrella term “Somatosensory Sensations” (Figure 1.1) and are accompanied by tactile, thermoreceptive and pain information, all conscious sensations arising from the periphery. Whereas *unconscious* proprioceptive senses may be referred to the proprioceptive senses contributing to automatic control of posture and movement, as well as joint stability.

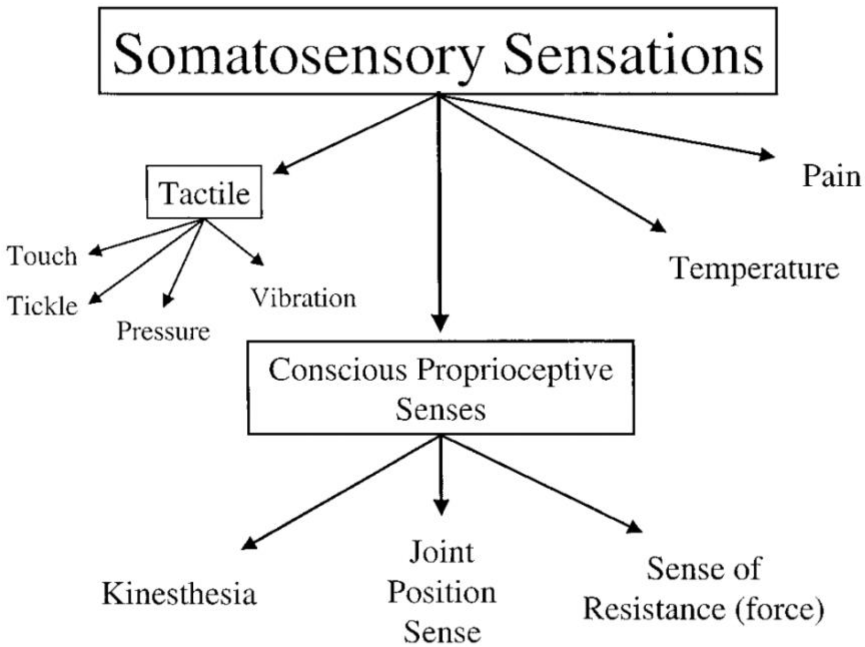


Figure 1.1: Sensations arising from somatosensory sources. (Riemann and Lephart, 2002).

Alternatively, Proske (2006) viewed the terms proprioception and kinaesthesia appropriate for addressing muscle afferents involved in *conscious* sensation only, while the term “muscle afferent input” was found to be more suitable in his opinion to describe *unconscious* automatic control of movement and posture. Although not opposing to the preceding researchers’ categorisation, it seems to distance unconscious sensation from any relation to proprioception. However, Proskes’ approach does echo Sherrington’s original definition of proprioception, where he goes on to describe it as “the perception of joint and body movement, as well as position of the body, or body segments, in space”. Here, perception may be defined as a complex, cognitive process used to interpret the *characteristics* or nature of the environment around us. Perhaps the clearest way of organizing proprioceptive terms is by dividing the proprioceptive system by function: sensory and non-sensory (Stillman, 2002). Sensory functions involve awareness of the spatial and mechanical status of the musculoskeletal framework, including—but not limited to—senses of position, movement and force. These "sensory functions" may collectively be termed "proprioception" or "proprioceptive sensation", and are integral to the development of motor control when learning new skills. Contrarily, the proprioceptive system's role in motor control during the performance of a learned skill is largely conveyed **without sensation**; as is its contribution to postural correction and reflex protection of the body against falls (balance) and of joints against harmful forces (automatic reflexes) (Stillman, 2002).

1.1.3 Types of proprioceptive sensation

The actual *process* of perceiving (in terms of proprioception) requires signals triggered by the stimulation of mechanoreceptors (receptors that relay proprioceptive and tactile information) to threshold through changes in any of the following (Proske and Gandevia, 2012):

-joint position

-joint movement

-force (including: resistance or tension, effort, and heaviness)

This in turn can be linked back to some degree to Sherrington's original description of the sub-modalities of "muscle sense": posture, passive movement, active movement, and resistance to movement (Riemann and Lephart, 2002).

In addition to these, a systematic review by Ager et al. (2020) classifies the sense of change in velocity (SoV), also known as the "sense of joint acceleration", as a subset of proprioception, separate from our sense of joint movement/ kinaesthesia. Upon closer inspection, it was found that SoV actually refers to our ability to detect vibration, attained by placing oscillating objects against ones' skin (Gilman, 2002; Shakoor et al., 2008). However, the context in which vibration (with regard to proprioception) has been discussed in the literature has generally been instrumental, used during joint matching experiments to elicit an illusion of movement and/or displaced position typically through vibration of the tendon or muscle (Gooey et al., 2000; White and Proske, 2008; Izumizaki et al., 2010). Moreover, definitions of vibration include *"a periodic back-and-forth motion of the particles of an elastic body or medium"* (Encyclopaedia Britannica, 2022) and *"a series of slight movements by a body back and forth or from side to side"* (Merriam-Webster, 2022). It may be argued that the sense of vibration should be classified as a subset of kinaesthesia rather than a type of proprioceptive sensation in and of itself, as vibration *is* a type of movement that so happens to occur around a joint (specifically on muscles or their tendons) in order to be sensed. On the other hand, when vibration is *sensed* (versus when a muscle tendon is vibrated to give an illusion of limb movement), it produces a very unique sensation different to that of joint movement. In addition, extensive studies on the human response to vibrations (in the context of automotive vehicle ride comfort) have established that the human body is sensitive to vibration frequency ranges from well below 1 Hz up to 100 kHz at the very least, which is a range of sensitivity even broader than our range of hearing, and which can illicit different responses in the human body, including motion sickness or kinetosis, and reduced coordination (Jeary et al., 1988). It has also been demonstrated in studies on roller coaster rides (Eager et al., 2016; Pendrill and Eager, 2020) that humans are able to sense both whole-body vibrations and local vibrations. Whole-body vibrations are typically transmitted to the human body through supporting surfaces such as a seat or floor, whereas local vibrations are experienced via a point of contact to the vibrating area, such as hand or foot transmitted vibrations. Therefore, it may be concluded that the sense of change in velocity should be classified as a separate type of proprioceptive sensation.

In addition to this, a robust systematic literature review by Horváth et al. (2022) included “trajectory sense” and “size sense” as part of the list of “aspects of proprioception” that may be assessed to accurately measure proprioceptive acuity. Although not as popular in the literature as other well-known proprioceptive sensations (joint or movement sense), it may be argued that each of these senses (trajectory and size) measure different aspects of proprioception than those of previously mentioned proprioceptive senses: for instance trajectory sense encompasses direction and shape of movement, while size sense requires one to assess the dimension of an object by gauging the space between ones’ digits simultaneously while holding the object.

All in all, the literature seems to indicate that there are six types of proprioceptive sensation as follows:

1. Joint (or limb) position sense (posture)
2. Kinaesthesia /Joint movement sense (passive or active movement)
3. Force sense, which includes: effort, tension and heaviness
4. Velocity sense (vibration)
5. Trajectory sense
6. Size sense

Joint position sense refers to a person's ability to perceive a given joint angle and reproduce it (passively or actively) after the limb has been moved to a different joint angle. Kinaesthesia or joint movement sense refers to the ability to recognise passive and active joint movement, be it the direction, amplitude or duration of this movement. The sense of effort and tension refer to the ability to appreciate force generated within the joint (Myers and Lephart, 2000a) and the sense of heaviness to judge the heaviness of objects (Proske, 2005). As for the sense of change in velocity, it refers to the ability to detect vibration, transmitted through supporting surfaces or points of contact with an oscillating object (Gilman, 2002). Trajectory sense, on the other hand, refers to the ability to actively reproduce the trajectory of a given movement. And lastly, size sense refers to the ability to discriminate between different sized objects (Horváth et al., 2022).

It is worth noting the use of the words *perceive, recognise, appreciate*, *judge*, *detect, reproduce, and discriminate* when defining the proprioceptive senses, as they all require a level of appraisal. This further supports that proprioception is a conscious process, distinct from, but not unrelated to, the subconscious level reflexes that correct posture as well as automate control of movement.

Additionally, proprioceptive sensations may be further divided based on the origin of sensory input. The senses of **effort**, **tension** and **heaviness** can be differentiated from other proprioceptive sensations in that they are always associated with sensory input *reafferent* in origin, being generated exclusively by motor command (Proske and Gandevia, 2012). This is not true with regard to joint position and movement senses however, as their sensory information can be exclusively *exafferent* in some instances, such as when a limb is moved passively in the absence of motor command. Yet they can also be perceived in the presence of motor command, accompanied by voluntary muscle contractions, for example when one moves their arm while supporting a load.

A valid question that may come to mind is: how does the brain differentiate between reafferent and exafferent signals when it comes to position and movement (or kinaesthetic) sensations? The hypothesis is that when self-generated and externally generated afferent activity is processed centrally, a copy of the motor command is sent to the anterior cerebellum. Here a comparison takes place between the expected spindle response based on past experience (also known as corollary discharge or efference copy) and the afferent signal coming from the muscle. Based on the match, an inhibitory signal is produced which supresses part or all of the reafferent component of the signal, preventing it from reaching the cerebral cortex so only the exafferent component is consciously perceived. This “cancellation” of the reafferent signal may also be referred to as “sensory prediction” (Wolpert and Flanagan, 2001) and it is thought that it provides the mechanism by which we are able to determine whether movement of our bodies has been self-generated or generated by an external factor, or perhaps by both in some cases. To that effect, our ability to identify the discrepancies between actual and predicted sensory feedback is integral in motor control, allowing us to generate appropriate motor behaviour under many various and often uncertain environmental conditions (Frith et al., 2000; Wolpert and Flanagan, 2001). And in turn, the development of this motor control, underpinned by proprioceptive awareness, helps us to acquire new skills (Stillman, 2002). This will be discussed in more detail under “**1.3 Anatomical and physiological representation of proprioception in the central nervous system”.**

1.1.4 Emerging literature on proprioception

Moving on to new and emerging literature on proprioceptive sensation, recent evidence (Proske and Chen, 2021) suggests there are actually two distinct senses of joint position that can be perceived simultaneously: one in which we are aware of limb position relative to our own body, also known as our "postural space", e.g. two-limb matching tests; and the other in which we are aware of limb position relative to the external world, also known as "extra-personal space", in which vision is a key component, e.g. when pointing or reporting verbally about limb position. This suggests that proprioception is multifaceted and may be sensed or perceived through various levels of consciousness. This is also corroborated by Heroux et al. (2022) in a perspective paper, in which they divide the assessment of human proprioception into two distinct categories: low-level and high-level proprioceptive judgements. Whereas low-level judgements refer to those made in a single frame of reference, such as joint matching or movement detection, high-level judgements involve those made in multiple frames of reference, such as when one is able to indicate the location of their limb in space. Our ability to produce high-level judgements indicates how instrumental proprioception is with regard to our sense of body ownership, or sense of self.

In conclusion, it appears that the study of proprioception is complex and still very much being dissected and explored. However, with what we know so far, **proprioception** may be defined as **a conscious *awareness* of the mechanical and spatial condition of the body through the integration of afferent sensory information arising from peripheral mechanoreceptors with previous sensory information.**

This “afferent sensory information” may be perceived in two ways: either **(1)** through the degree of change in muscle length and/or tension, joint angle, and/or stretch of skin; or **(2)** as a result of external stimuli, such as visual and tactile stimuli. Defined in this way, proprioception is differentiated from the *unconscious, automatic reflexes* involved in motor control, which require recruitment of the same mechanoreceptors (e.g. muscle spindles and Golgi tendon organs) to control movement, but do not require direct awareness or conscious perception of the mechanical and spatial condition of the body.

Moreover, the integration of this “afferent sensory information” with “previous sensory information” refers to the process of judging current sensory stimulation with anticipated or predicted feedback generated from previous similar sensory experiences. Taken together, this sensory information shapes the “proprioceptive acuity” of an individual through learning and memory, and is critical to postural and motor control, contributing to one's sense of body ownership.

1.2 Receptors of Proprioception

A mechanoreceptor is a type of specialised sensory receptor which converts mechanical events (deforming the host tissue) into frequency-modulated neural signals which travel through afferent sensory pathways to the central nervous system (CNS) for processing. Mechanoreceptors which are responsible for the generation of proprioceptive information can be classified into three categories: muscle receptors, joint receptors and cutaneous receptors.

The mechanoreceptors primarily contributing to proprioceptive information are thought to be located in muscles, tendons, joint capsules, and ligaments, with those situated in the fascia and deep skin layers considered supplementary sources of information (Proske, 2005).

On a side note, it is worth mentioning that while reviewing the literature on proprioception, it was noticed that most researchers seem to prefer using the term "receptors of proprioception" rather than "proprioceptors" when describing these specialised mechanoreceptors. The reasoning for this, as it seems, is the contribution of these receptors not only to conscious sensation, which—as mentioned previously—is believed to be a condition of proprioceptive sensation (Proske and Gandevia, 2012), but also to "unconscious" automatic reflexes of motor control. This reiterates the idea that proprioception, although related to the automatic control of movement in the body, is ultimately a conscious process by definition.

Having said that, the three categories of mechanoreceptors responsible for proprioception within the body can be further classified as follows:

1) Muscle receptors:

-*Muscle spindles* (Figure 1.2) embedded within skeletal muscles are a type of stretch receptor, responsible for conveying information regarding change in muscle length and rate of changes in length (velocity) (Riemann and Lephart, 2002) related to joint position and movement, and potentially play a role in sensing force (Proske and Gandevia, 2012).

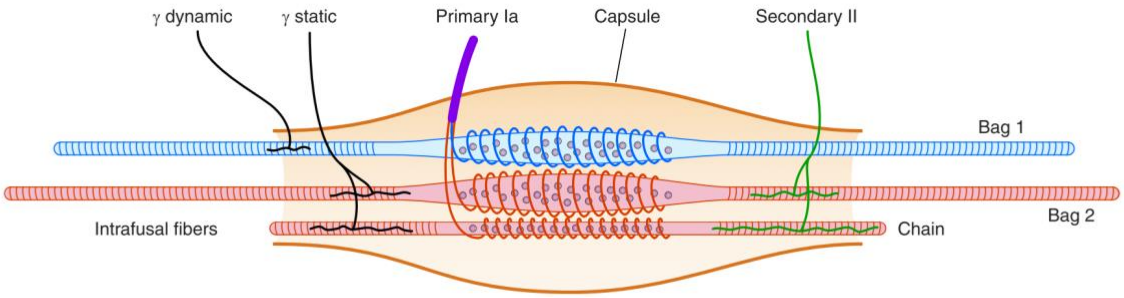


Figure 1.2: A diagrammatic representation of the muscle spindle. Shown are the intrafusal fibres, which include the large nuclear bag 1 (in blue) and bag 2 (in red) fibres, in addition to the smaller nuclear chain fibres (also in red). The ends of the chain fibres lie within the capsule, while the ends of the bag fibres extend beyond it. All three intrafusal fibre types are bound (around their nucleated portions) with large, group Ia afferent fibres (purple) which terminate as primary endings onto these fibres. While smaller, group II afferent fibres (in green) terminate as secondary endings, located to one side of the primary endings, and supplying bag 2 and chain fibres. Gamma static fusimotor fibres also innervate bag 2 and chain fibres, while gamma dynamic fusimotor fibres only innervate bag 1 fibres (Proske and Gandevia, 2012).

-*Golgi tendon organs* (Figure 1.3 and 1.4), which are located at the interface of muscles and tendons (the myotendinous junction), are also stretch receptors. However, they are—for the most part—a contraction receptor, sensing primarily active muscle tension. Most tendon organs can be considered regional tension sensors, as they can signal tension in a select group of motor units; also, they may provide a calibrating signal for the sense of effort (Riemann and Lephart, 2002; Proske, 2005; Proske and Gandevia, 2012).

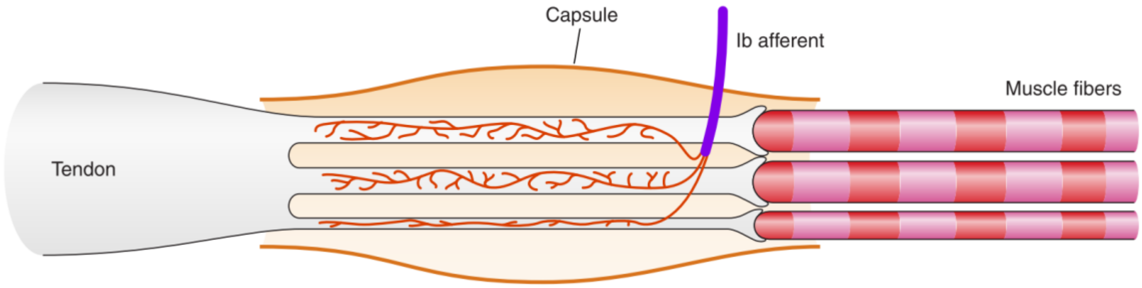


Figure 1.3: A diagrammatic representation of the Golgi tendon organ in mammals. The receptor capsule is penetrated by a group Ib axon, which then branches out with each branch terminating on a tendon strand attached to a muscle fibre. Typically, one tendon organ can have 10^3 muscle fibres attached to it, with each fibre a part of a different motor unit. When one of the motor units supplying a tendon organ contracts, the tendon strand attached to the corresponding motor unit stretches, which generates activity in the Ib axon (Proske and Gandevia, 2012).

2) Joint receptors:

Embedded in the joint capsules (Figure 1.4), these receptors consist of low threshold sensory receptors typically associated with the tactile system such as *Ruffini endings* and *Pacinian corpuscles*. These receptors' responses often peak at extreme ranges of joint position, suggesting they play a role as "limit detectors" within the joint (Tuthill and Azim, 2018). Moreover, although some evidence suggests joint receptors may play a role in the sense of limb movement (but not position), joint position studies indicate that the role of joint receptors in position *and* movement sensation is not significant (Proske and Gandevia, 2012; Proske, 2015a).

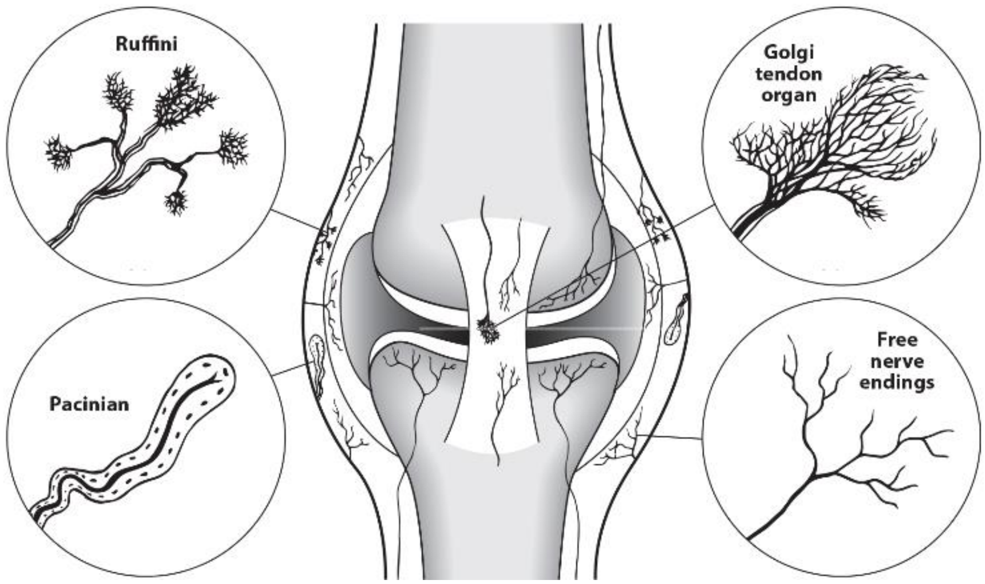


Figure 1.4: A diagrammatic representation of Ruffini endings and Pacinian corpuscles. These tactile receptors are embedded in the joint capsule here, making them joint receptors in this case. Also shown are the Golgi tendon organ(above right) embedded in the myotendinous junction; and free nerve endings, also known as nociceptors, which are a type of sensory receptor that detects noxious stimuli, such as extremes in temperature, pain, and damage to the tissue. Illustration by Zimny and Wink (1991).

3) Cutaneous or Tactile receptors:

The literature commonly refers to four primary types of cutaneous receptors (Figure 1.5): Two located superficially in the epidermis: slowly adapting type I, *Merkel's discs,* and rapidly adapting type I, *Meissner's corpuscles*; and two located deeper in the dermis: slowly adapting type II, *Ruffini endings,* and rapidly adapting type II, *Pacinian corpuscles* (Riemann and Lephart, 2002).

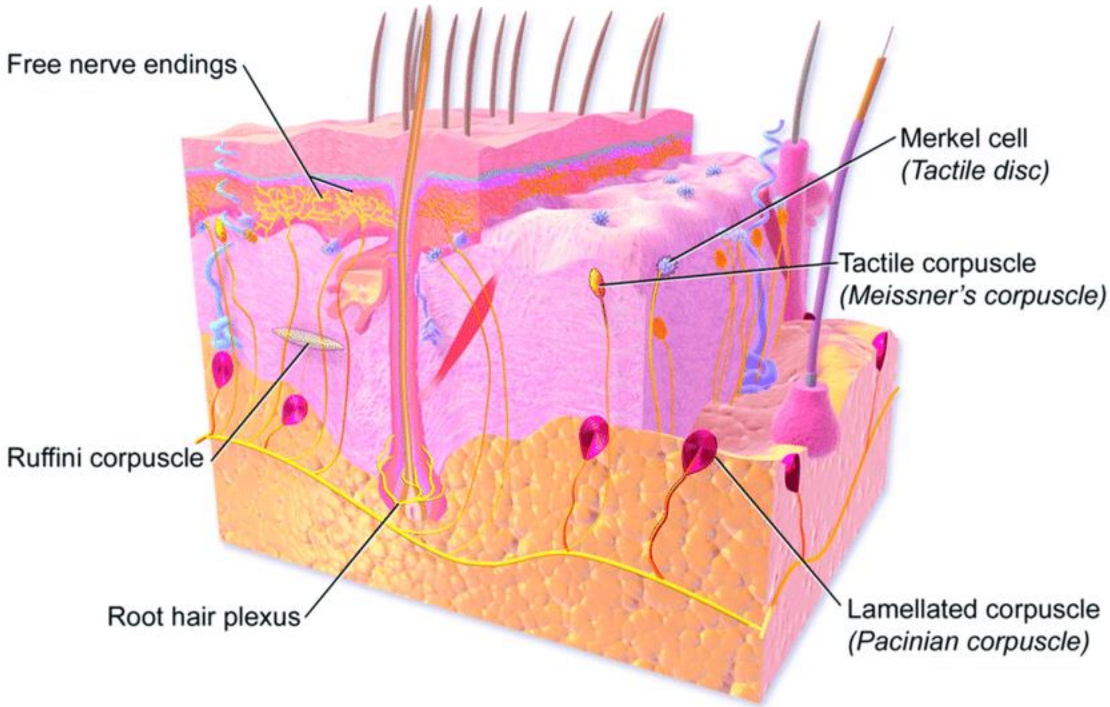


Figure 1.5: A diagrammatic representation of cutaneous receptors. Merkel's discs and Meissner's corpuscles reside in the epidermis, while Ruffini endings and Pacinian corpuscles are located in the dermis of the skin. Illustration by Xie et al. (2017).

As mentioned previously, these receptors typically play more than the role of receptors of proprioception in the body. Muscle spindles and Golgi tendon organs, for example, are also involved in unconscious, reflex control of movement (Proske and Gandevia, 2012).

Additionally, cutaneous receptors may act both as receptors of proprioception and exteroceptors while the body is in motion. For example, sensations of elbow movement may be perceived as a result of skin stretching over the elbow joint, which is proprioceptive. However, when our elbows come into contact with external objects in the environment, cutaneous receptors act as exteroceptors to register the sensations (Proske, 2015).

Although the role of different mechanoreceptors in the formation of a proprioceptive experience has been debated widely in the literature, there is unanimous support of the idea that muscle spindles (Figure 1.2) are primarily responsible for our proprioceptive abilities (Ribeiro and Oliveira, 2011; Proske and Gandevia, 2012). Experiments involving thixotropic conditioning as well as vibration of passive muscles are among the strongest indicators corroborating the idea of muscle spindles' primary role in proprioception:

-**Thixotropy experiments:** Thixotropic conditioning of muscles involves manipulating spindle background activity through (1) previous contraction and relaxation of the muscle without changing its length (isometric contraction) or with a change in length (isotonic contraction: concentric/shortening or eccentric/lengthening) or (2) by passively holding the muscle in a lengthened or shortened position, before testing for joint position (Macefield and Knellwolf, 2018). Data from research on the effect of muscle conditioning on position sense at the forearm (Allen et al., 2007) and ankle (Reynolds et al., 2020) confirms a significant change in perceived limb position following muscle conditioning. The underlying idea is that when muscle spindles (stretch receptors) are stretched, there is a direct simultaneously proportional increase in their discharge rate. The brain interprets a higher rate of discharge as a *longer* muscle, yet what is perceived is not a longer muscle but a more flexed or extended joint. This misinterpretation leads to joint matching errors in the absence of vision (Allen et al., 2007). Because spindle background activity is believed to be a component of the spindle response in charge of signalling limb position (Proske, 2015), these experiments indicate that muscle spindles play an integral role in the perception of joint position.

-Vibration experiments: In an experiment by Ribot-Ciscar et al. (1998) sustained vibration applied to the passive muscle tendons of the ankle muscles was found to decrease resting spindle firing rate, in addition to reducing spindle sensitivity to passive stretch immediately after the vibration was stopped. These changes in spindle activity are believed to account for the illusions of limb movement and displaced position produced, even after vibration of the corresponding muscle tendon has ceased, and can last up to 40 seconds before full recovery. In addition, in an experiment by Allen and Proske (2006) on the effect of muscle fatigue on kinaesthesia, vibration of the elbow flexor tendon was performed to confirm movement detection was indeed coming from muscle spindles during trials of movement tracking. It was found that subjects perceived the movement as being faster than it actually was when the reference arm was vibrated, and slower when the indicator arm was vibrated.

Underpinned by numerous other experiments (Wise et al., 1996; Gregory JE, Morgan DL, Proske U, 2004; Allen et al., 2007), these phenomena are regarded as the most influential in support of muscle spindles as the principal receptor of proprioception. In addition, observational data on patients who have undergone joint replacement surgeries, as well as patients who have sustained dorsal column lesions, also support this concept. For example, it was found that patients who have had total hip replacement surgery, which involves removal of all capsular and ligamentous tissue, retained both their position and movement senses entirely (Proske and Gandevia, 2012). These proprioceptive senses were also shown to be preserved in the lower limbs of patients who had sustained damage to the dorsal column of the thoracic spine, as the muscle afferents (of the lower limbs) leave the dorsal column below this level, specifically at the upper lumbar spine, and thus remain intact. However, skin and joint afferents do take a pathway through the dorsal column at the thoracic region and, by implication, are affected by dorsal column lesions, with extensive loss of skin and joint sensation in the legs observed clinically. This also indicates that skin and joint input is actually not required for relatively normal position and movement sense in the legs (Proske and Gandevia, 2012).

In addition, it has been established from experiments on animals that muscle spindles are particularly dense in muscles where *especially accurate* control of movement is important, such as in the intrinsic muscles of the hands (in bonnet monkeys) and in the neck (in cats) (Macefield and Knellwolf, 2018). Human experiments have indicated an abundance of muscle spindles in the neck region as well, likely indicating they play an important role in integrating proprioceptive information of the head with respect to the body (Macefield and Knellwolf, 2018).

1.3 Anatomical and physiological representation of proprioception in the central nervous system

1.3.1 The three levels of motor control

Sensory information received from mechanoreceptors are integrated at three distinct levels of motor control in the central nervous system (CNS): at the spinal level, at the brain stem, and at the higher levels of the CNS such as the cerebral cortex and cerebellum (Ribeiro and Oliveira, 2011). Each of these levels generate unique motor responses crucial to controlled, coordinated movement, as well as functional joint stability. At the spinal cord, descending commands from the brain stem and cortex can control axons conveying proprioceptive information via neurons and interneurons connecting with higher CNS levels. This indicates that supraspinal areas of the CNS also play a role in the *modulation* of proprioceptive information that enters the ascending tracts. Most proprioceptive information travels via two distinct pathways to the supraspinal regions of the CNS:

-the dorsal lateral tracts, which convey information to the somatosensory cortex

-the spinocerebellar tracts, which terminate, as the name implies, in the cerebellum

While the fastest signal transmission velocities in the CNS—which are associated with unconscious, automatic motor performance—are exhibited by the spinocerebellar tracts, it is the dorsal lateral tracts that are responsible for conscious proprioceptive performance (Riemann and Lephart, 2002).

In terms of functionality, each of the three levels of motor control mentioned previously contribute in regulating the way we move: At the spinal level, direct motor responses—in the form of reflexes—contribute to functional joint stability. At the brain stem level, in order to control automatic and stereotypical movement patterns, in addition to balance and postural control, afferent information is integrated with visual and vestibular input. However, it is at the higher regions of the CNS, specifically the cerebral cortex and the cerebellum, where conscious awareness of proprioception is generated, contributing to movement refinement and gesture diversification (Myers and Lephart, 2000; Velasques et al., 2013).

Furthermore, during a task-oriented activity, the integration of sensory input (including proprioceptive input) from the three levels of the CNS facilitates coordination and stability of the body in anticipation of movement execution (also known as a feedforward command), in addition to correcting for velocity and timing errors *during* its execution (feedback command) by comparing actual movement with intended or predicted movement (Vega et al., 2021). It is believed that this predicted movement is derived from past motor events (see Figure 1.6). This process of modifying movement based on error feedback is otherwise known as motor adaptation (Vega et al., 2021) and is necessary in coping with any changes occurring in the external and internal environment. It also promotes motor learning by updating the internal feedforward motor command (Gelener et al., 2021).

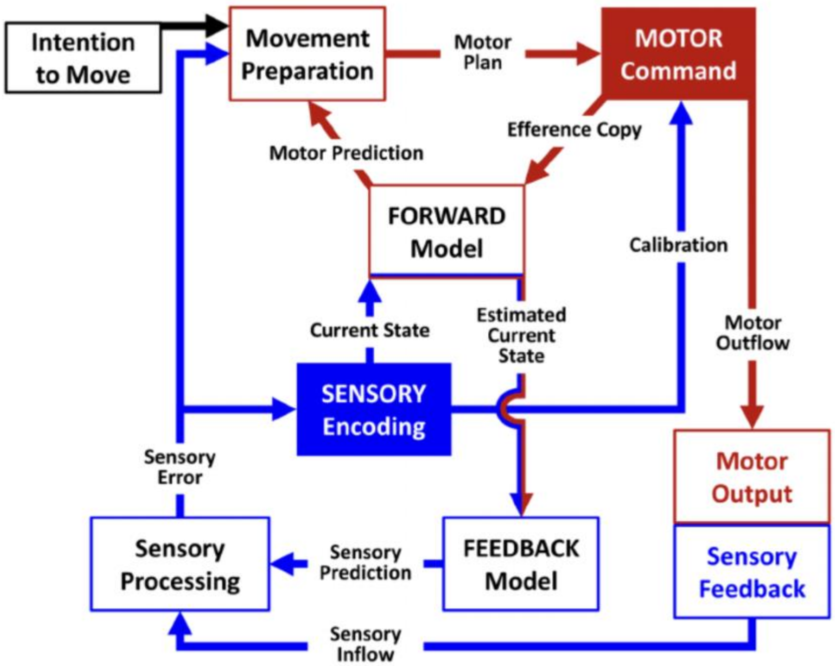


Figure 1.6: A diagrammatic representation of sensory-motor integration. This specific diagram is based on the Sensory-Motor Integrative Loop for Enacting (SMILE) model theory. The motor command signal (red arrows) incorporates an efference copy which has previously been processed by the forward model and a motor outflow generating a sensory feedback (blue arrows). A comparison of actual sensory feedback and anticipated or predicted feedback (generated by a feedback model) is made and information on the resulting sensory error is transmitted to higher level regions of the CNS in order to calibrate the subsequent motor command (Perruchoud et al., 2014).

1.3.2 Motor learning

Perhaps the importance of proprioception is most evident during the learning of new skills, where an acute awareness of postures and movements is required. This in turn leads to a refinement in movement patterns, specifically fine motor skills. For example, when learning how to type, there is high consciousness particularly of wrist, hand and finger movements; this allows the individual to more readily change their movement patterns to be more efficient. As learning proceeds and typing movements become more refined, afferent feedback signals from the respective body segments are systemically stored in the brain as templates of properly executed typing movements (efference copies). Research using functional magnetic resonance imaging and positron emission tomography indicate that the main sites for this learning process are the prefrontal cortex and the cerebellum (Stillman, 2002). Once fully learned, typing demands minimum use of proprioceptive consciousness of the corresponding body segments, and maximum use of stored efference copies; so at a subconscious level afferent signals fed back from proprioceptive receptors in the periphery are cross-checked against the stored efference copies to verify correct performance. This shifts control of learned activity from a conscious cerebral process to a largely subconscious one; which allows the focus to be more on the outcome of the activity rather than the actual process (Stillman, 2002). This reiterates the point made previously about learned skill being conveyed without sensation: Although the learning of the skill requires tremendous proprioceptive awareness, once it is learned we are able to utilise it subconsciously, almost as if on autopilot.

1.4 Proprioception and postural control

It may be deduced from the above (Figure 1.6) that accurate proprioceptive input from both external and internal sources facilitates successful preparation and controlled maintenance of the body and/or it's segments during the execution of task-oriented activities. This "controlled maintenance" of body position during daily activities may otherwise be defined as **postural**

control; and in reality, it is the interaction of *several* sensory-motor processes that effectively lead to postural control (Gelener et al., 2021). These include:

-Movement strategies, such as reactive, anticipatory and voluntary movement strategies

-Sensory strategies in the form of sensory integration and sensory reweighting

-Cognitive processing, such as learning and memory

-Orientation in space, for example perception of visual or postural verticality

-Bio-mechanical constraints, such as strength, degrees of freedom and base of support

The coordination of these processes to maintain postural stability during both intrinsic and extrinsic disturbances results in <u>postural equilibrium</u> as well as <u>postural orientation</u>. While postural equilibrium is needed to support the body during static (i.e., quiet standing) and dynamic (i.e., reaching and gait) activities of daily living, postural orientation involves positioning of the body with respect to gravity, supporting surfaces, and the visual environment, among other sensory reference frames (Gelener et al., 2021).

Moreover, the term “postural control” in itself is often (and preferably) used when quantitative data is needed concerning human balance; in other words, human balance may be re-defined as the “clinically observable outcome” of the postural control system in humans (Şimşek and Şimşek, 2020).

Having said that, it has been shown that balance is one of the functional abilities most affected by proprioceptive impairment (Fling et al., 2014; Labanca et al., 2021).

Additionally, there is general agreement in the literature that signals arising from receptors of proprioception located in lower limb muscles provide the primary source of information for postural control, and that those arising from the ankle region specifically are the most important in terms of balance (Goble et al., 2011; Fling et al., 2014; Han et al., 2015; Henry and Baudry, 2019). Several studies on clinical populations corroborate this idea: A descriptive study investigating the effect of sensory deficits on balance, functional status and trunk control in patients (11 male and 9 female) with Guillain-Barré syndrome (GBS) found a significant correlation between ankle proprioception (measured via joint position sense) and standing balance (p=0.023, 0.022 for the right and left ankles respectively) whereas the effect of knee proprioception on standing balance was not significant (p>0.05) (Huzmeli et al., 2018). Also, a recent cross-sectional study on 57 patients (42 male and 15 female) with chronic symptoms following stroke found that ankle proprioception (measured via both movement and joint position sense) was the strongest predictor of balance impairments in patients who demonstrated balance problems after stroke (p=0.025) (Cho and Kim, 2021). In addition, a retrospective study (Butler et al., 2014) on 234 total joint arthroplasty patients (75 hips, 65 knees and 94 ankles) examined single leg balance approximately one year following surgery. Patients who were able to maintain balance on a single leg for 10 seconds with their eyes open were considered successful at the test. It was found that total hip (63%) and total knee (69%) arthroplasty patients had similar success rates, whereas total ankle arthroplasty patients showed significantly lower success rates (9%) (p< .05). This seems to indicate the importance of the ankle joint specifically in maintaining balance.

Moreover, it is thought that proprioception of the ankle joint is of particular importance during quiet standing, as receptors at the ankle region provide the most salient information with regard to body sway, by detecting minute variations in muscle length produced by the rotational movement around the ankle joint (Henry and Baudry, 2019). Furthermore, several studies have confirmed that the disruption of proprioceptive signals during standing leads to increased body sway (Inglis et al., 1994; Lajoie et al., 1996; Horak et al., 2009); while other studies have estimated that proprioception alone contributes up to 58% of body sway performance during standing (Fling et al., 2014).

Given that body sway is a well-accepted indicator of postural stability/balance, it may be deduced from the above that ankle proprioception is not only related to standing balance; it also seems to be closely linked to body sway during standing.

1.5 Clinical relevance of balance impairments

The importance of good standing balance is that it directly impacts our ability to perform activities of daily living (ADLs)(Cho and Kim, 2021). Therefore it is no surprise that inability to balance during standing can have life changing implications: Along with decreased functionality during ADL's, it has been shown that quality of life can be negatively impacted in people with balance impairments following stroke (Schmid et al., 2013; Azadeh et al., 2018) and in those with mild to moderate Parkinson's disease (Kurt et al., 2017).

Moreover, balance disorders are believed to be one of the main causes of falls in older adults (Bednarczuk and Rutkowska, 2022). Generally speaking, falls have been shown to be a major healthcare issue worldwide: Annually, approximately 37.3 million falls that occur are severe enough to require medical attention, and an estimated 684 thousand falls lead to death, making it the second leading cause of unintentional injury death after road traffic injuries (World Health Organization, 2021). However, it is the demographic age group of older adults suffering the greatest percentage of falls: Approximately 28-35% of people aged 65 years and over fall each year, with an increase of up to 42% for those over the age of 70 years; multiple falls per year also occur in > 15% of people aged over 65 and 25% of those over 80 years (World Health Organization, 2008). In addition, even when non-lethal, falls among the elderly can often lead to life debilitating injuries, including hip fractures or traumatic brain injury (Quijoux et al., 2019).

Therefore, it would seem that developing rehabilitation programs that focus on strategies to decrease falls, or more precisely the risk of falls, should be among the priorities when treating patients with balance disorders.

1.6 Studies on standing balance and proprioception

A recent study (Bednarczuk and Rutkowska, 2022) examined the risk of falls in active older female students (n=36, mean age 67.11 ± 5.35 years) by testing their standing stability during 3 different conditions: double leg stance with eyes open (1) and eyes closed (2) on a stable platform, and Falls Risk Test (3) on an unstable platform. It was found that the overall stability index calculated in measurements with eyes closed predicted the risk of falls most accurately in the older active females (R^2=0.391, F(1.34)=23.475; p<0.000). These results seem to indicate that tests conducted with eyes closed, (when the ability to stabilise oneself relies mostly on proprioceptive and vestibular feedback without any visual cues) are the most sensitive in determining balance disorders in physically active women, further solidifying the idea that proprioception and standing balance are closely related. The study also indicates that preventative programmes should include exercises performed with eyes closed, in order to challenge proprioceptive capability and develop it so as to increase standing stability.

With regard to the ankle specifically, a randomised-controlled, single blind study (Karakaya et al., 2015) investigated the relationship between static balance and ankle proprioception. The study included 59 healthy university students (35 females, 24 males) with ages ranging from 18-28 years, and these subjects were divided into experimental (n= 29) and control (n=30) groups. The experimental group received an “ankle proprioceptive” exercise program, which included stretching and strengthening of the ankle joint muscles, as well as balance board exercises 5 days a week for 2 weeks for a total of 10 sessions. The control group received no intervention. At the end of the program, static body balance was evaluated using a kinaesthetic ability trainer (a specialised force plate used with software to assess standing balance using specific balance index scores) for both unipedal and bipedal stance. Although balance scores improved for both groups, the experimental group showed significantly lower index scores than the control group ($p<0.05$), indicating better ability to balance. It was concluded that ankle proprioceptive training had positive effects on static body balance parameters in healthy individuals. However, it may not be entirely accurate to label the training program used in this case as proprioceptive:

there was no mention of any exercises done with the eyes closed for instance, nor was there any explanation about how the ankle training program actually engaged proprioceptive sensations, such as is the case when performing proprioceptive tests. However, this vague classification of "proprioceptive training" of the ankle joint is somewhat common in the literature. For example, a high-quality systematic review and meta-analysis which summarised the available evidence on the effectiveness of proprioceptive training in preventing ankle sprains (Schiftan et al., 2015) assessed 7 randomised controlled trials, all of which used challenging balance exercises for their interventions: these included standing on one leg with eyes closed or with eyes open while completing a task, or standing on an unstable surface made especially for training such as a wobble board or ankle disc, or a combination of the above. While it may be appreciated that the aforementioned training programs were indeed effective in reducing the incidence rate of ankle sprains in the athletic population, it seems inaccurate to call the programs "proprioceptive" training. Rather, they should be referred to as balance training, which encompasses visual and vestibular feedback in addition to that which is proprioceptive. However, there is also convincing empirical evidence that balance interventions induce large gains in proprioceptive function (Beckerle et al., 2022). This raises a very valid question: We know that balance and proprioception are closely related, and we know that proprioception is a multifaceted, conscious sensation; so which facet of proprioception can best predict balance ability?

1.7 Proprioception testing methods

In order to answer the question above, a brief description of the three **main** testing techniques (based on classic psychophysical methods) used for assessing proprioception according to (Han et al., 2016) follows:

1. **Threshold to detection of passive motion (TTDPM)** measures a persons' ability to detect passive movement of the corresponding body segment of various joints across the body (Figure 1.7 A-C). Typically, the clinician or investigator controls the motion by way of a machine, with predetermined directions and speeds. During a TTDPM assessment, the body site being tested is typically isolated by strapping adjacent body segments, and other sensory information such as visual, tactile and aural information is blocked using the appropriate equipment. As soon as motion is detected, participants press a button to report they have detected the motion. Scores may be based on time taken—or amplitude of movement—needed to detect motion.

2. **Joint position reproduction (JPR)**, also referred to as joint position matching (JPM), can be conducted either passively or actively, and may involve either ipsilateral or contralateral limb movements. There are 3 types of JPR tests described in the literature: ipsilateral JPR (which requires the use of memory to reproduce the joint position of the same limb), and 2 contralateral JPR approaches, one in which memory is required, the other in which it is not due to the contralateral limb being held in the reference position during the JPR test. See Figure (1.7 D-F) for examples of JPR testing.

3. **Active movement extent discrimination assessment (AMEDA)** tests are performed using active movements on an AMEDA apparatus (Figure 1.7 G-I). Following a familiarisation trial, in which a number of movement displacements of differing distances are experienced actively by the subject, the testing begins: Subjects are presented with one of the several displacement positions experienced previously by way of allowing them to move the tested limb to a physical "stop", followed by a return to start position. Subjects are then asked to make a judgement as to the position number (also learned during the familiarisation trial) of each test displacement using their memory and without any feedback.

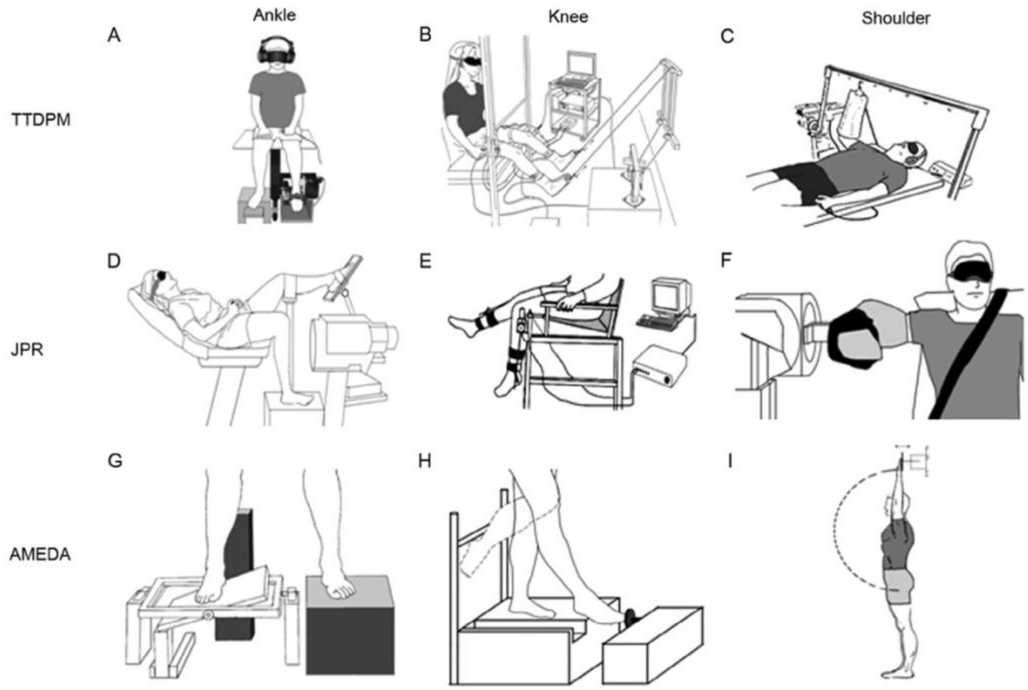


Figure 1.7: Illustrations of the three main proprioception testing techniques. Threshold to detection of passive motion (TTDPM) A-C, joint position reproduction (JPR) D-F, and active movement extent discrimination assessment (AMEDA) G-I (Han et al., 2016).

Each of these tests have been developed from different concepts, are performed under different testing conditions, and are thought to assess different aspects of proprioception (Han et al., 2016; Yang et al., 2020). However, because of this fact, the tests are not directly comparable (Krewer et al., 2016), meaning they are not alternative methods, rather they are complimentary to each other. Additionally, the AMEDA test method in particular has received various criticisms (Krewer et al., 2016; Héroux et al., 2022). Although it is thought that AMEDA does indeed assess movement discrimination, it is argued that information from various sensory systems at multiple joints is available and required during testing, i.e. while standing to perform the test, visual and vestibular information, along with proprioceptive information from various joints and body segments, are required to maintain standing position. This indicates that the results obtained by AMEDA are not measuring ankle proprioception alone. Having said that, this paper will continue to review literature that has applied the AMEDA method to assess ankle proprioception, keeping in mind that results may not be indicative solely of ankle proprioceptive performance.

Another thing to consider is the seeming lack of any research on tests conducted to assess the senses of force, velocity, trajectory or size in the literature. The reasoning for this seems to be the inconsistent uptake of these constructs in rehabilitation settings at present (Hillier et al., 2015). For the most part it appears that rehabilitation programs are focusing on the assessment of kinaesthetic senses (position and movement), which is reflected in current literature. A more detailed discussion about proprioception testing methods will follow in the methods chapter.

1.8 Linking balance and proprioception: A literature review

Various studies have used these proprioceptive tests in an effort to understand the relation between proprioception and balance in light of other factors such as ageing and neural disorders. The following review will divide studies to those conducted on healthy participants of all ages, and those carried out on clinical populations:

1.8.1 Balance and proprioception studies: Healthy populations

With regard to healthy populations, a study by Deshpande et al. (2016) evaluated the association of proprioception at the ankle joint with objective, as well as self-reported, measures of balance, mobility and physical performance across the adult life span. A total of 790 healthy participants (362 females) with an age range spanning from 24-97 years were divided into 4 groups as follows: n<40 years = 20, <60 years = 128, <80 years = 436, and ≥80 years = 206. Outcome measures included TTDPM to measure ankle joint proprioception, and for balance, mobility and physical performance, single leg stance, gait speed, and the short physical performance battery (SPPB) were used as objective measures, while subjective outcome measures included the walking index, in addition to being asked questions about fear of falling and mobility difficulties. Performance was divided into quintiles according to TTDPM scores with the lowest TTDPM scores (best proprioceptive performance) in the 1st quintile and the highest TTDPM scores (worst proprioceptive performance) in the 5^{th} quintile. Results indicated that ankle proprioception, as measured by TTDPM, was significantly associated with age. A significant, graded main effect of age group was found where the two older age groups were significantly different from the two younger age groups (p <.001–.016) and from each other (p<.001). More importantly, it was found that participants in higher TTDPM quintiles demonstrated significantly worse performance across the board for measures of balance, in addition to mobility and physical function (p<.001-.05).

Another study using the TTDPM method to measure proprioception was by Song et al. (2021), who explored the relationship between proprioception, cutaneous sensitivity and muscle strength with balance control among healthy older adults. A total of 164 participants (89 females, mean age 73.5 ± 7.8 years) were tested for proprioception at their dominant knee and ankle joints, along with other tests measuring cutaneous sensitivity and muscle strength of the ankle region. To measure balance, the Berg Balance Scale (BBS) and the root mean square (RMS) of the centre of pressure (COP) were used as indications of static and dynamic balance control. Results showed a significant—albeit weak to moderate—correlation between ankle proprioception and balance, where proprioception during ankle plantar flexion (PF) ($r = -.306$, $p = .002$) and dorsiflexion (DF) ($r = -.217$, $p = .030$) were weakly to moderately correlated with BBS, and proprioception during ankle DF ($r = .220$, $p = .035$) was weakly correlated with COP-RMS in the medio-lateral direction. It should be noted that both of these studies used only one method of proprioceptive assessment (TTDPM) as the lone measure of ankle proprioception, and so results may not be generalisable to ankle proprioception per se. Because performance in different proprioceptive tests has not been shown to be well correlated (de Jong et al., 2005; Yang et al., 2020) general proprioceptive status should not be inferred from the assessment of a single proprioceptive method. Had either of the other methods been used (JPR or AMEDA) results may have been quite different. It was also noticed that participants of both studies were tested in a seated position during the TTDPM assessment, which according to Refshauge and Fitzpatrick (1995) is not the ideal position for testing TTDPM as it has been shown that subjects are able to perceive comparatively smaller ankle movements (a third of the size) when they are either upright in a standing position or when seated with knees and ankles extended versus when they are seated with the knees bent and the ankles in PF. The reasoning behind this seems to be an enhanced input from muscle spindles when the calf muscle is stretched during knee extension and ankle DF (Refshauge and Fitzpatrick, 1995).

Moreover, a retrospective study by Huang et al. (2017) aimed at investigating whether measures of balance correlate with measures of proprioception, in this case both TTDPM and JPR, used existing data of 80 healthy young participants (43 female, mean age 31 ± 10.4 years). Three measures of balance (star excursion balance test (reach)- response to perturbation (in seconds) and number of foot lifts during single-leg stance) were tested for correlation with ankle inversion/eversion TTDPM and JPR proprioceptive measures. Proprioception during ankle inversion specifically has been suggested as a more useful measure of proprioception than during plantar flexion in healthy populations (Teasdale et al., 2017) as “proprioception” has been shown to be increased in joint ranges in which there is greater use, resulting in a practice effect for proprioception.

However, for this study (Huang et al., 2017) only one negative correlation was found, between TTDPM and single leg stance foot lifts ($r=-.30$, $p=.008$). This may suggest that the chosen balance assessment methods were unable to capture the relationship between balance and ankle proprioception. In addition, it may have been useful if ankle DF and PF data were available to analyse against the chosen balance assessments, especially given the influence ankle DF and PF musculature seem to have on body sway (Warnica et al., 2014; Yoshizawa and Yoshida, 2022).

On that note, a recent study by Shin and Kim (2022) looked for correlations between balance (measured using body sway with either eyes open (EO) or eyes closed (EC)) and JPR for the lower extremities (hip, knee and ankle), among other outcome measures. A total of 30 young healthy university students (mean age 21.5 ± 1 years) participated in the experiment. With regard to ankle proprioception, results showed that ankle PF JPR error was significantly correlated with sway length EO ($r=.76$, $p<.05$), as well as sway length EC ($r=.77$, $p<.05$). It was concluded that for the proprioception outcome measure, ankle joint PF JPR error was the best predictor of sway length. However, the choice of sway length as a measure of balance may not have been ideal, as it has been shown repeatedly that sway *velocity* tends to show the smallest reproducibility error across trials and populations and is able to discriminate well between test situations (Raymakers et al., 2005; Sakanaka et al., 2021).

1.8.2 Balance and proprioception studies: Clinical populations

Moving on to clinical populations, a recent cross-sectional study by Cho and Kim (2021) aimed to determine the main factor that predicts balance impairments in patients with chronic stroke. The study consisted of 57 patients (42 males, mean age 55.7 ± 12.2 years) with chronic symptoms following stroke, who were tested for several ankle function outcome measures, including proprioception (TTDPM and JPR methods) and balance (BBS and Timed-Up and Go (TUG) tests). To assess proprioception, participants were seated with knees flexed at 90° and fixed to only allow ankle movement. They were asked to place their affected foot on the footplate of the ankle movement device, and to place their non-affected foot on the height matched footrest. After a familiarisation session, the proprioception assessments were carried out in 2 consecutive steps: first the affected ankle was moved passively from neutral (0°) to randomly assigned 10° target angles (10°, 20°, or 30° for ankle PF and inversion; 10° or 20° for ankle DF and eversion) while asking the participant whether they detected the movement and direction of movement. After staying at the target position for 5 seconds, the ankle was returned to the starting position (0°). For the second step, the affected ankle was passively moved again toward a target angle, and the participant was asked to report when they felt that they had reached the target angle. No feedback about results was provided to the participant during testing. Three ankle movements were evaluated per direction, with a total of 38 measurements, including blank trials, performed. As for the balance assessments, the BBS and TUG tests classified 21 of the patients as having a balance impairment. Results indicated that ankle proprioception, as measured by TTDPM and JPR, was the strongest predictor of balance impairment for stroke patients with balance problems ($p<.05$). However, caution should be exercised with proprioceptive tests using ipsilateral matching tasks, such as was performed in this study, when participants are among clinical populations (stroke in this case). According to Goble (2010), it is likely in this situation that some portion of matching error measured actually reflects cognitive or memory deficits, rather than any decrease in proprioceptive performance.

Another recent study investigating proprioception within clinical populations was by Chen et al. (2020), who explored the relationship between joint position sense of the lower extremities and motor function in children with developmental coordination disorder versus typically developing children. A total of 56 participants were recruited; 28 children with developmental coordination disorder (15 males, mean age 10.86 ± 1.07 years) and 28 typically developing children (16 males, mean age 10.96 ± 1.18 years). Joint position sense, assessed using the JPR method (passive ipsilateral) with an isokinetic dynamometer (Biodex), was measured at the knee and ankle, and position error (PE) as well as position error variability (PEV) were determined. As for motor function, it was examined using the 2nd edition of the Movement Assessment Battery for children (MABC-2) and quantified using sub-scores from 3 MABC-2 domains. Results showed that PE and PEV for both knee and ankle joints were significantly greater (both $p<0.01$) in children with developmental coordination disorder compared with typically developing children. More importantly, ankle joint PE and PEV were significantly greater than knee joint PE and PEV (both $p<0.01$) in the children with developmental coordination disorder. Also, there was a significant negative correlation between joint position acuity in the lower extremities and the MABC-2 balance sub-score ($p<.05$). These results indicate that not only do children with developmental coordination disorder exhibit poorer ankle JPR than knee JPR, but their PE and PEV scores tended to predict balance function, i.e. greater PE and PEV were related to lower balance scores.

As for the AMEDA method for testing proprioception, two studies were found which utilise it to examine the relationship (albeit indirectly) between ankle proprioception and balance. Both studies examined ankle proprioception in people with Parkinson's disease (PD). The first of the two was by Teasdale et al. (2017) who conducted a cross-sectional observational study including 13 participants with mild to moderate PD (mean age 71 ± 31 years) and 14 healthy age-matched controls (mean age 66 ± 21 years). The study aimed to determine whether people with PD have impaired proprioception in the ankles during active movement, and if there are correlations between ankle proprioception and fear of falling, history of falls, and Parkinsonian symptoms. As mentioned before, the AMEDA method was used to assess ankle proprioception. The other outcome measures included The Falls Efficacy Scale (FES), Parkinson's Disease Questionnaire (PDQ-39), and self-reported number of falls during the previous 12 months. Although results showed that people with PD had significantly worse proprioception (as measured by AMEDA) in plantarflexion, inversion, and overall proprioception than control participants ($p<.05$ all), there were no significant relationships between impaired proprioception and fear of falling or fall history. There was, however, a moderate negative correlation between impaired proprioception and Parkinson's symptoms $r=0.441$, $p=.021$). Again, because the AMEDA method is a multi-sensory test, it is possible that participants made use of other sensory feedback to discriminate between test movements, such as cutaneous feedback from the sole of the foot, or from the auditory cues from the AMEDA device itself. The second of the two studies was conducted recently by Wang et al. (2021), and examined whether ankle proprioception and functional mobility were related in people with PD. A total of 42 participants with mild to moderate PD volunteered and were tested for ankle proprioception using the AMEDA method while standing, as well as for functional mobility using the Timed-Up and Go (TUG) test, a type of dynamic balance test, the 30 second Sit to Stand (30s-STS) test and the 10-meter walking (10MWT) test. Again, in this study no significant correlations were found between ankle proprioception and functional mobility performance measures in people with PD ($p>.05$), except for moderate correlations between ankle proprioception and step length ($r=0.38$, $p<.05$) as well as step cadence ($r= -0.30$, $p<0.05$). The authors justify these results by suggesting that people with Parkinson's rely on other sensory input to achieve functional mobility, consistent with sensory reweighting theory. However, it may just be that the proprioceptive assessment method AMEDA allows participants in general (with PD and without) to rely on other feedback rather than ankle proprioception to perform the test. A similar test to AMEDA was used in a study by Deshpande et al. (2003), called "active-to-active reproduction of joint position". In the study, which aimed to determine reliability and validity of several ankle proprioceptive measures, participants were positioned into standing with support provided by a backrest with Velcro belts at the waist, hip, knee and foot levels. For the active-to-active JPR the participants closed their eyes during testing to eliminate visual input. During testing, participants actively moved their ankle through their available range of motion at a self-selected speed and then stopped at the test target positions on the experimenters' instructions. They remained in this position for 5 seconds, after which they moved the ankle back to the starting position. Participants were then asked to reproduce the test positions actively and data was collected after participants indicated that they had reproduced the target position.

The absolute difference between the test and reproduced position was recorded as the position error. Although this method has not been as commonly referred to in the literature as the AMEDA method, it seems better able to capture ankle proprioceptive performance for two reasons: first, although participants are positioned in standing, they are able to rest their body weight on the backrest, allowing them to maintain standing with minimal muscle activity. This in turn may minimise feedback from hip and knee joints, allowing sole concentration during testing to be on the ankle joint. Second, there were no physical stops used in active-to-active JPR as there are in AMEDA, which may allow participants to utilise mostly their ankle joint position sense as well as memory of joint position during the test; whereas with the physical stops, one may use other subtle types of feedback in addition to these such as auditory cues as well as cutaneous feedback when reaching the physical stops.

Table (1) summarises the findings of our literature review for a clearer picture of the proprioceptive testing methods used for each study and whether the findings of these studies indicated any particular proprioceptive testing method to be a good predictor of balance ability.

Author/s	**Year**	**Study title**	**Studied population**	**Balance testing method**	**Proprioceptive testing method**	**Main findings**	**Did proprioceptive test significantly predict balance ability?**
Deshpande et al.	2016	Ankle proprioceptive acuity is associated with objective as well as self-report measures of balance, mobility, and physical function.	Healthy adults (n=790) aged between 24-97 years	Single-leg stance, gait speed, the short physical performance battery (SPPB) and general mobility and physical performance	TTDPM	-High TTDPM scores at the ankle were associated with worse balance performance. -Significant effect of age	Yes

						of TTDPM scores.	
Song et al.	2021	Relationship of proprioception, cutaneous sensitivity, and muscle strength with the balance control among older adults.	Healthy older adults (n=164) aged 73.5 ± 7.8 years	Berg balance scale (BBS) and root mean square of centre of pressure (COP)	TTDPM	-High TTDPM scores during both dorsi and plantar flexion of the ankle associated with worse balance performance	Yes
Huang et al.	2017	Measures of balance are not correlated with proprioception at the ankle joint.	Healthy young adults (n=80) aged 31 ± 10.4 years	Star excursion balance test, response to perturbation, and number of foot lifts during single-leg stance	TTDPM and JPR	-Only TTDPM was found to be associated with balance performance.	Yes for TTDPM No for JPR
Shin and Kim	2022	Relationships among Lower Extremity Muscle Circumference, Proprioception, ROM, Muscle Strength, and Balance Control Ability in Young Adults.	Healthy young adults (n=30) aged 21.5 ± 1 years	Body sway COP length	JPR	-Ankle plantar flexion highly correlated with balance measured via COP sway length	Yes
Cho and Kim	2021	Ankle Proprioception Deficit Is the Strongest Factor Predicting Balance Impairment in Patients With Chronic Stroke.	Stroke survivors (n=57) aged 55.7 ± 12.2 years	BBS and timed-up and go (TUG) test	TTDPM and JPR	-Ankle proprioception found to be the strongest predictor of balance impairments in stroke patients	Yes for both
Chen et al.	2020	Joint position sense of lower extremities is impaired and correlated with balance function in children with developmental coordination disorder.	Children with developmental coordination disorder (n=28) aged 10.86 ± 1.1 years and age matched healthy children (n=28) aged 10.96 ± 1.2 years	Movement assessment battery for children (second edition)	JPR	Ankle joint positional error (PE) was shown to be a better predictor of balance function than knee PE	Yes
Teasdale et al.	2017	Proprioception of the Ankle is Impaired in People with Parkinson's Disease	Patients with mild to moderate Parkinson's disease (PD) (n=13) aged 71 ± 31 years and age matched healthy adults (n=14) aged 66 ± 21 years	Falls efficacy scale, Parkinson's disease questionnaire, and self-reported number of falls in 12 months	AMEDA	-Proprioceptive performance at the ankle was worse for PD patients compared to controls. -Proprioceptive performance correlated to Parkinson symptoms	No

Wang et al.	2021	The Relationship Between Ankle Proprioception and Functional Mobility in People With Parkinson's Disease: A Cross-Sectional Study.	Patients with mild to moderate PD (n=42)-ages not disclosed	TUG test, 30 second sit-to-stand test and the 10-meter walking test	AMEDA	-Ankle proprioception correlated with step length and cadence.	No

Table 1.1: A summary of our literature review comparing proprioceptive assessment methods with balance tests. Studies on healthy participants (first four studies from the top) indicate that assessing proprioception using the TTDPM method seems to predict balance performance quite well, whereas JPR was not as consistent. However, for the clinical populations, both TTDPM and JPR were able to significantly predict balance performance. Although no studies could be found comparing ankle proprioception measured via the AMEDA method with balance performance in healthy populations, two studies on patients with Parkinson's disease indicated that no correlations were found between ankle proprioception using the AMEDA method and balance performance.

Abbreviations: Short physical performance battery (SPPB), Threshold to detection of passive motion (TTDPM), Berg balance scale (BBS), Centre of pressure (COP), Joint position reproduction (JPR), Timed-up and go (TUG), Parkinson's disease (PD), Active movement extent discrimination assessment (AMEDA).

We may conclude from our review that TTDPM seems to be the most common method used to assess ankle proprioceptive acuity in general, and seems to be a good predictor of balance ability in healthy and clinical populations. However, only two studies out of 8 compared more than one method of proprioceptive assessment, and both of these studies presented with several methodological issues, such as inappropriate balance assessment methods (Huang et al., 2017) and using methods to measure proprioception that involve cognitive function for patients with stroke (Cho and Kim, 2021). Another limitation of the above studies is the lack of a standardised measure for balance ability. Only two of the studies reviewed use a gold standard measure of standing balance (Song et al., 2021; Shin and Kim, 2022), and neither of these studies assessed more than one method of proprioception, therefore it is difficult to compare results between these and the other studies. When it comes to proprioceptive experiments, lack of a specific framework or protocol to follow when choosing assessment methods only adds to the difficulty of generalising the findings of these studies. We believe that adhering to a specific outcome measure or tool to assess balance—and we nominate body sway in this case given it is a well-accepted indicator of balance ability—would be a step in the right direction. Furthermore, it remains unknown which measure of proprioception best predicts balance ability using a single method of balance assessment.

1.9 The effect of muscle strength on proprioception

While reviewing the literature related to balance and proprioception, several studies we came across examined the effect of muscle strength and exercise on proprioceptive acuity (Tsang and Hui-Chan, 2003; Butler et al., 2008; Ribeiro and Oliveira, 2010; Guney et al., 2016; Wang et al., 2016; Niespodziński et al., 2018). Given the location of arguably the most influential receptor of proprioception—the muscle spindle—(Ribeiro and Oliveira, 2011; Proske and Gandevia, 2012) is within skeletal muscles, it seems appropriate to study the effect of muscle strength on proprioceptive acuity while analysing the relationship between proprioception and balance. At the very least, we hope to add additional dimension and depth to our findings, whether significant or not.

1.10 Conclusion

It may be concluded from the literature review that although various studies have explored the correlation between balance and proprioception, there still seems to be a gap in the literature on which proprioception assessment method best predicts balance ability. Answering this question would have direct clinical implications on the elderly community specifically, a population known to have a high incidence rate of falls which often lead to life debilitating injuries. Moreover, developing accurate methods to assess and treat balance impairments is often a high priority goal of neurorehabilitation, due to the direct effect it can have on patients' ability to perform activities of daily living and—just as importantly—on their psychological well-being and quality of life. Another consideration that should be taken into account is the practicality of the proprioceptive assessment method used: ideally it should be concise and relatively easy to perform as to not induce fatigue, as the target clinical populations whom which these tests will be carried out upon will likely fatigue easily.

The current study attempts to answer the following research question: "**Which proprioception assessment method best predicts balance ability in healthy young and elderly populations?".** This question is underpinned with the aim of developing proprioceptive testing methods that are practical in terms of time and effort needed to complete them, as it is the intention of the researcher to implement the developed testing methods on populations with balance impairments in the future if possible. To the knowledge of this researcher, no other study has indicated which proprioceptive testing method correlates best with balance ability—using one standard method to measure balance ability—while keeping in mind the need for practicality while testing to enable the recruitment of those more prone to fatigue such as elderly individuals and those with balance impairments.

We hypothesised that one of the proprioceptive assessment methods presented would have a moderate to strong correlation with standing balance measured via body sway in healthy adults.

Experimental hypothesis: There will be a moderate to strong correlation between at least one of the proprioception testing methods used and standing balance measured via body sway in healthy adults ($p<.05$).

Null Hypothesis: There will be no correlation between at least one of the proprioceptive testing methods and standing balance measured via body sway in healthy adults ($p>.05$).

To begin testing our hypotheses, a methodology chapter explains and discusses the different assessment methods developed and performed to assess proprioception, specifically at the ankle joint. Then, the proprioception testing methods are compared and contrasted experimentally to find the one most suitable in terms of time and ease of application. Last but not least, balance of healthy adults is measured and analysed against proprioception of the ankle to look for any relationships between the two. We also attempted to analyse the effect of muscle strength on proprioceptive acuity within the scope of our experiments.

Answering our research question requires a dissection and exploration of different methods of proprioceptive assessment. Therefore, a brief description of the aim of each of our experimental chapters follows:

Our first experimental chapter "Chapter 3: The Effect of Age and Strength upon Joint Position Reproduction" aimed to determine whether muscle strength had an effect on the age-related decline in proprioceptive acuity. We measured proprioceptive performance through a matching task, one of the proprioceptive methods used in our research, and determined whether participants' strength had an effect on this proprioceptive performance. Our second experimental chapter "Chapter 4: Active and Passive Detection of Threshold" aimed to determine the effect of isometric contraction on proprioceptive acuity at the ankle. We compared proprioceptive performance through another method of proprioceptive assessment—known as detection of threshold—while the calf-muscle was either in an active or passive state. Our third experimental chapter "Chapter 5: Proprioceptive Reaction Time as a Predictor of Body Sway" aimed to determine whether proprioceptive acuity at the ankle predicted body sway performance during quiet standing. Using a third proprioceptive assessment method, known as proprioceptive reaction time, we looked for a relation between proprioceptive performance at the ankle joint and balance performance measured through body sway centre of pressure.

While our overall aim was to answer our research question "**Which proprioception assessment method best predicts balance ability in healthy young and elderly populations?"**, the studies we performed did not fully meet the overarching research aim stated above. Our initial planned programme of research involved comparing several different aspects of proprioceptive acuity–as measured through at least three different proprioceptive assessment methods–to quiet standing balance measured through body sway. This would entail adding **joint position reproduction** and **amplitude threshold detection** assessments to the **reaction time** assessment in our last experiment (Chapter 5) to compare the proprioceptive performance of our participants during each of these tests with their quiet standing balance performance. We would then be able to determine which of these proprioceptive assessment methods best predicted standing balance. However, we ended up comparing only reaction time performance to quiet standing balance. This was due to time constraints as a result of unforeseen circumstances, including the Covid 19 pandemic, as well as several personal medical events.

Chapter 2: Methods

Abstract

For our methods chapter, we go into a detailed presentation of the aspects of proprioception found in current literature. These are then divided based on the psychophysical tests used to measure proprioception, which include the method of limits, the method of adjustment, and the method of constant stimuli. A table with a summary of our findings is also presented. As there was an appropriate joint position reproduction(JPR) method already in place to test proprioceptive acuity at the ankle, we aimed to develop a method to measure threshold to detection of passive motion (TTDPM). We then go on to describe the process of developing an appropriate proprioceptive testing method. We discuss the thinking process, sequence of experimentation and detailed procedures used, and conclude with the resulting two methods developed. Both methods developed fall under the "method of limits" psychophysical branch of testing. The first method, in which we incorporate a psychophysical staircase to obtain threshold scores, used angular amplitude (measured in degrees) as the main outcome measure. This method was adequate for use on healthy young participants, but it was decided the method was too time consuming were it to be used for older-aged populations, or populations with balance deficits. We then developed our second method, in which reaction time (measured in seconds) was the main outcome measure. This method proved to be less time consuming and was shown to be appropriate for use on healthy older-aged participants.

2.1 Introduction

Measuring proprioception is rarely a straight-forward task. As established in the introduction section, proprioceptive sensation is a multifaceted process, linking lower level autonomic responses with higher level conscious processes in order to not only control movement and posture but to refine it to a level in which movement patterns are perfected. Consequently, it would seem that assessing proprioceptive acuity cannot be captured using a single measurement method. In addition to this, it appears from the literature review that no widely accepted standard assessment of proprioception exists. Having said that, objective measurements are an important component of clinical assessment, allowing professionals in the field to create bench marks of functional ability for their patients. These "bench marks", more formally referred to as reference points, allow clinicians to establish specific, measurable and functional goals as part of their patients' rehabilitation program. In relation to balance deficits, evaluation of proprioceptive sense is typically an important element when outlining a treatment plan with the aim of improving overall motor control (Gelener et al., 2021) due to the significant link between balance and proprioception previously discussed in the literature review.

Moreover, clinicians and researchers have developed many methods to capture proprioceptive acuity, in order to investigate the impact of proprioception loss on function in clinical populations most implicated by it: these include stroke survivors, those recovering from age-related falls, and peripheral neuropathy, in addition to those with movement disorders such as Parkinson's and Huntington's disease (Hillier et al., 2015). Two studies in particular clearly describe a set number of methods to measure proprioceptive accuracy; the first was by Hillier et al. (2015) who identified 3 criterions of methods: joint position detection, passive motion detection threshold, and passive motion direction discrimination. The other, by Han et al. (2016) described 3 paradigms as well (which were described briefly in Chapter 1), however, they were based on the three classic psychophysical methods of assessment: (a) the method of limits, in which participants are asked to indicate when the appearance (or disappearance) of a stimulus is perceived, and which the threshold to detection of passive motion (TTDPM) test is based on, (b) the method of adjustment, in which participants are asked to adjust the level of stimulus to a reference, and which the joint position reproduction (JPR) test is based on, and lastly (c) the method of constant stimuli, in which stimuli are presented in pairs and participants are asked to compare them, and which is the basis of the active movement extent discrimination assessment (AMEDA). Although both reviews add much needed clarity to those navigating the available proprioceptive research literature, it seems that they share an important limitation which is their—somewhat—narrow definition of proprioception. A recent review by Horváth et al. (2022) applies a more inclusive approach to proprioception than the previous reviews, taking into account the growing literature on proprioceptive testing methods, including new tests which have been developed since the previous reviews' publication. Similar to Han et al. (2016), Horváth et al. (2022) used the classic psychophysical methods mentioned above to categorise the various types of proprioceptive tests (Table 2.1). Typically, psychophysics are defined as a quantitative method of analysing the relationship between a physical stimulus and the sensations or perceptions produced as a result of this physical stimulus, making it a sensible tool when attempting to quantify proprioceptive acuity. A detailed explanation of each of the psychophysical methods will follow.

Aspect of proprioception	Method of limits	Method of adjustment	Method of constant stimuli
Joint position sense	Threshold to detection of passive motion	Joint position reproduction	Joint position discrimination
Movement sense		Movement reproduction	Movement discrimination
Force sense	N/A	Force reproduction	Force discrimination
Muscle tension sense	N/A	Muscle tension reproduction	N/A
Weight sense	N/A	N/A	Weight discrimination
Velocity sense	N/A	Velocity reproduction	Velocity discrimination
Trajectory sense	N/A	Trajectory reproduction	N/A
Size sense	N/A	N/A	Size discrimination

Table 2.1: Summary table of proprioceptive accuracy test methods. Adapted from (Horváth et al., 2022).

N/A: No method available

2.2 Psychophysical methods for measuring proprioception

2.2.1 Method of Limits

In this method, the stimulus may be presented in either an ascending or descending order. For the ascending method, the experimenter presents the participant with a barely detectable stimulus, which is gradually increased until it is reported to be perceived by the participant. For the descending method, this process is reversed: the level of stimulus is decreased until it is no longer perceived. For both methods, the threshold is the point where the stimulus is just detected. Typically, both ascending and descending methods are alternately used in the same experiment and their thresholds are then averaged.

The disadvantage of this method is that participants may be prone to habituation, or conversely, anticipation errors. The **staircase method**, which is an adaptive psychophysics method of testing, may be used to avoid these errors (Geschieder, 1997; Leek, 2001).

The staircase method was introduced by Georg Von Békésy in 1960 during an auditory perception experiment (Geschieder, 1997). Staircases typically begin with a high-intensity stimulus, which can easily be detected. The stimulus intensity is then reduced gradually until it is reported to be undetectable by the subject, at which point the staircase is reversed, with the stimulus increased in steps until the subject is able to detect the stimulus once more, which triggers another reverse/change in direction. This method allows the researcher to pinpoint the detection or perception threshold for the given stimulus by averaging the last values of these "points of reversal" (Leek, 2001).

With regard to proprioception, the **method of limits** may be used to measure threshold to **detection** or **perception** of passive motion (TTDPM, TTPPM respectively, and used interchangeably). A TTDPM test can be performed in standing, sitting or lying down, and typically all other sensory information, such as visual, tactile, and auditory information, is blocked out using equipment such as blindfolds or goggles, air-cushions, and noise-cancelling headphones (Han et al., 2016). The limb is positioned into a device to hold it into place and to stabilize the adjacent joints, and is then passively moved into a predetermined direction, at a predetermined speed. It has been shown during TTDPM testing that the velocity of movement has an effect on the individuals' ability to detect the actual movement, and it has been demonstrated that the slower the velocity, the higher the threshold, meaning the more difficult it is to detect motion (Refshauge and Fitzpatrick, 1995). Taking this into account, along with the staircase method of measurement, the process of threshold detection may be streamlined in the lab to be less time and energy consuming. After the limb is passively moved, the subject is instructed to press a button as soon as movement and direction are detected. If the direction reported by the subject is incorrect, the trial is discarded and testing continues until 3 to 5 correct answers are acquired. This is in line with the methodology for TTDPM testing used by Nagai et al. (2012) in which it was determined that 3-5 correct trials of TTDPM resulted in good reliability and precision, with an intra-class correlation coefficient (ICC) of (0.72 to 0.86) and (0.68 to 0.85) for intra-session and inter-session reliability analyses, respectively. As such, this methodology will be one of a few adopted for our current research to measure proprioceptive threshold of the ankle joint, and will be presented in more detail further along in the chapter.

2.2.2 Method of Adjustment (also known as Method of Average Error)

In this method, a predetermined level of stimulus is introduced either passively (by the experimenter) or actively (by the participant) for a set amount of time, after which the stimulus is switched off or set back to the start. The subject is then asked to adjust the level of stimulus (variable stimulus) so that it is perceived by them to be as close to the predetermined "reference" stimulus as possible.

In terms of proprioception, the **method of adjustment** can be used for measuring several aspects of proprioceptive accuracy related to reproduction of a stimulus, as presented in Table (2.1). For our research, we will focus on joint position reproduction (JPR), as it can be measured reliably and swiftly with equipment already available in our lab. During JPR, either ipsilateral or contralateral joints may be used: For ipsilateral joint position matching (IJPR), the specified joint is placed into the target/reference position, either actively or passively, and is then returned to the start position. On that note, some evidence suggests that matching accuracy is improved when reference position is established through the participants' own active movement rather than when it is determined passively by the experimenter (Goble, 2010). It is thought that during voluntary (active) limb movements, better sensory information from muscle spindles in particular result in smaller positional errors compared with passive limb movements (Hung and Darling, 2012; Han et al., 2016).

Referring back to IJPR, after establishing the reference joint position, subjects are then asked to reproduce the reference joint position using the same limb, a task which requires the use of memory. As for contralateral joint position matching (CJPR), it may be conducted in two ways: the first is identical to the IJPR method in experiencing the reference position after which the limb is returned to the start position; however, the subject is then asked to reproduce the reference position using the **contralateral** limb, which, like IJPR, also requires **memory** of the reference position. The second way to measure CJPR is by allowing the limb to remain at the reference position and asking the subject to move the contralateral limb into what they perceive to be as close to the reference joint position as possible. This would require no memory of the reference position, as the subject has "live" ongoing information about the joint position to allow them to replicate it on the contralateral side. The difference measured between the reference position and the adjusted one is considered to be the "joint position error", and over several tests it is averaged to get the "mean error", considered a measure of the participants' sensitivity to the stimulus (Han et al., 2016). This mean error may also be interpreted as a measure of accuracy, with the standard deviation of this error a measure of precision.

2.2.3 Method of Constant Stimuli

For this method, sensory threshold is determined by presenting a set of stimuli repeatedly, where the level of stimulus intensity is in no particular order (random). So instead of being presented in ascending or descending order, similar to the method of limits, levels of the stimulus are not related from one trial to the next. This prevents subjects from predicting the intensity of the next stimulus, reducing errors of expectation and habituation. Furthermore, the threshold in this method is defined as the stimulus intensity or value that yields a detection response 50% of the time (Binder et al., 2009).

There are two type of thresholds that can be obtained using the method of constant stimuli (Geschieder, 1997; Han et al., 2016):

1. Absolute threshold: The lowest possible **intensity** of a stimulus that one can detect
2. Difference threshold: The lowest possible **difference in intensity** that one can detect between two stimuli being compared simultaneously, with one of the stimuli being the standard stimulus and the other being the comparison or variable stimulus

Although thought by some researchers as the most accurate of the three psychophysical assessment methods for studying the psychophysics of movement (Han et al., 2016), it should be noted that a very large number of trials would be required to obtain an acceptable measure of threshold using this type of test, which may not be very practical, especially in a clinical setting (Geschieder, 1997). To tackle this issue, the active movement extent discrimination assessment (AMEDA) method was developed and has been validated for testing proprioception acuity of the shoulder, hand, cervical and lumbar spine, hips, and knees (Han et al., 2016). Despite the growing popularity of the AMEDA method, it has gained criticism in recent years for its methodology (Krewer et al., 2016; Héroux et al., 2022) and specifically it's construct validity with regard to ankle proprioception: It is thought that results obtained by AMEDA while assessing ankle proprioception do not represent ankle proprioceptive acuity per se, rather they are more representative of a "multi-joint measure of a multi-segment posture". In addition, it is argued that a person could theoretically successfully perform the AMEDA test without accessing ankle proprioception at all by using cutaneous and auditory feedback while performing the test (Héroux et al., 2022).

Although it was the intention of the researcher to apply the AMEDA method as part of the ankle proprioceptive assessments for this dissertation, time constraints did not allow for this to take place, and given the recent controversy, it may be just as well.

Nevertheless, with regard to proprioception in general, the **method of constant stimuli** may be used to measure our ability to discriminate between various stimuli such as joint position and movement, force, weight, velocity and size (Table 2.1).

2.3 Developing an appropriate proprioceptive testing method

2.3.1 Thinking process

In order to answer our research question "*Which proprioception assessment method best predicts balance ability in healthy young and elderly populations?"*, we needed to develop and compare various tests of ankle proprioception in terms of time needed to complete tests from start to finish, and ease of completing the tests, in an attempt to minimise fatigue as much as possible specifically for the older subjects participating.

However, we first wanted to establish if there were any associations between ankle proprioception and muscular ability at the ankle joint, particularly as our experiments would involve measuring standing balance via body sway, a task that requires coactivation of the lower limb musculature, with emphasis on the triceps surae muscle. We carried out two separate studies to complement our research: the **first** exploring the relationship between muscle *strength* and proprioceptive acuity, particularly among older individuals, to determine whether age-related decline in proprioception is affected by changes in muscle strength; the **second** aimed to compare (a) proprioceptive acuity during active muscle contraction of the triceps surae with (b) proprioceptive acuity during the passive state of muscle tension (of the same muscles) during rest. These experiments would in turn provide insight for developing the assessment methods for our final study, as well as inform future treatment programs and protocols that may be used clinically in balance programs for older-aged populations in particular.

2.3.2 Sequence of experiments

We began our research with a joint position reproduction (JPR) experiment (Chapter 3) using an experimental methodology already utilised for previous research by our predecessors. We had several aims from this initial experiment, the first of which was to explore the effect ageing had on one's ability to reproduce a specific angle at the ankle joint, secondly to examine whether a persons' muscle strength showed any correlation with their proprioceptive acuity, and thirdly to investigate whether muscle strength had any effect on the decline in proprioceptive acuity due to ageing.

Following this initial experiment, we went on to develop a new experimental method to measure proprioceptive acuity using threshold to detection of passive motion (TTDPM). The piloting process undertaken to refine this method will be the subject of the remainder of this current chapter, where we will go on to describe the sequential development of the methodology used for the subsequent experiments. After this initial piloting, four experiments using the TTDPM method were performed for our research, the last in which we attempted to find any correlations between proprioceptive acuity using reaction time and body sway. Although it was our intention to perform additional experiments, following the ones we completed, to look for correlations between other proprioceptive acuity tests (e.g. JPR and amplitude TTDPM) and body sway, time constraints did not allow these plans to see fruition. A summary of the sequence of our experimentation, numbered from first to last experiment is presented below:

1. The Effect of Age and Strength upon Joint Position Reproduction:
   Presented in Chapter 3, we used a contralateral joint position reproduction (JPR) set-up which was in place for our predecessor researchers during their experimentation.
2. Threshold to Detection of Passive Motion methods development:
   This process will be presented and discussed as part of the current chapter (sections 2.4 and 2.5) starting with pre-piloting and piloting, after which we demonstrate the various adaptations that were made to produce the final

version of our experimental methods. The flowchart (Figure 2.1) below illustrates a summary of this process:

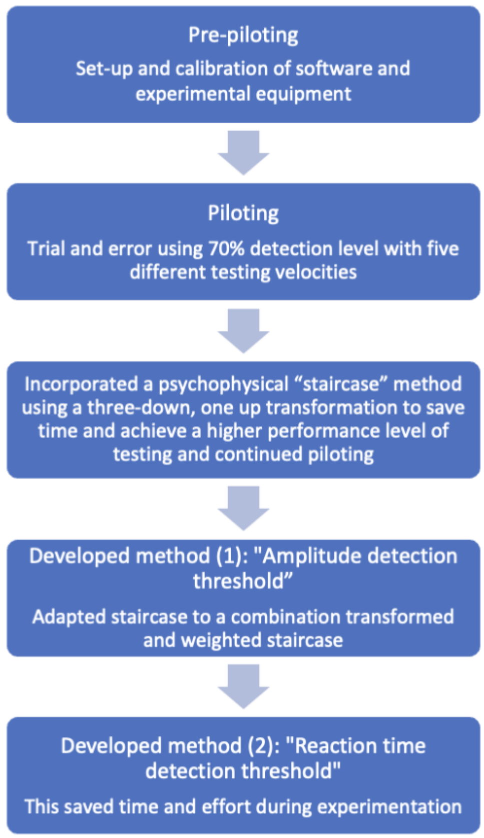


Figure 2.1: A flowchart depicting the sequence of our methods development. After establishing our first method we went on to perform two experiments, after which it was decided that a different, less time-consuming method was needed. Our second method, using reaction time as the main outcome measure was then developed, and two additional experiments were carried out.

3. Amplitude detection threshold among different velocities: Presented in Chapter 4, this was the first experiment conducted using our new amplitude detection threshold method. It was a preliminary study from which we aimed to establish the average detection threshold scores among participants.

4. **4.** Active and Passive Detection of Threshold: Also presented in Chapter 4, we conducted this experiment following the previous preliminary one. It was during this experiment that we realised the testing method would be too time consuming—and fatigue-inducing—for our older participants. We went on to develop our second method, reaction time threshold, which was shown to be markedly less time consuming than the previous method.
5. **5.** Reaction time threshold among different velocities: Presented in Chapter 5, we conducted another preliminary study using the reaction time method to—again—establish average threshold scores among our participants.
6. **6.** Proprioceptive Reaction Time as a Predictor of Body Sway: Following the preliminary study, we conducted our last study (also presented in Chapter 5). This experiment was the largest among our experiments in terms of recruitment number, and we were able to successfully assess older aged individuals in a quick and comfortable manner using the reaction time method. We then assessed participants body sway to analyse the relationship between proprioception as measured via reaction time and balance as measured via body sway.

<u>2.3.3 Experimental procedure</u>

For all the experiments carried out, ankle proprioception was assessed using the apparatus depicted in Figure (2.2):

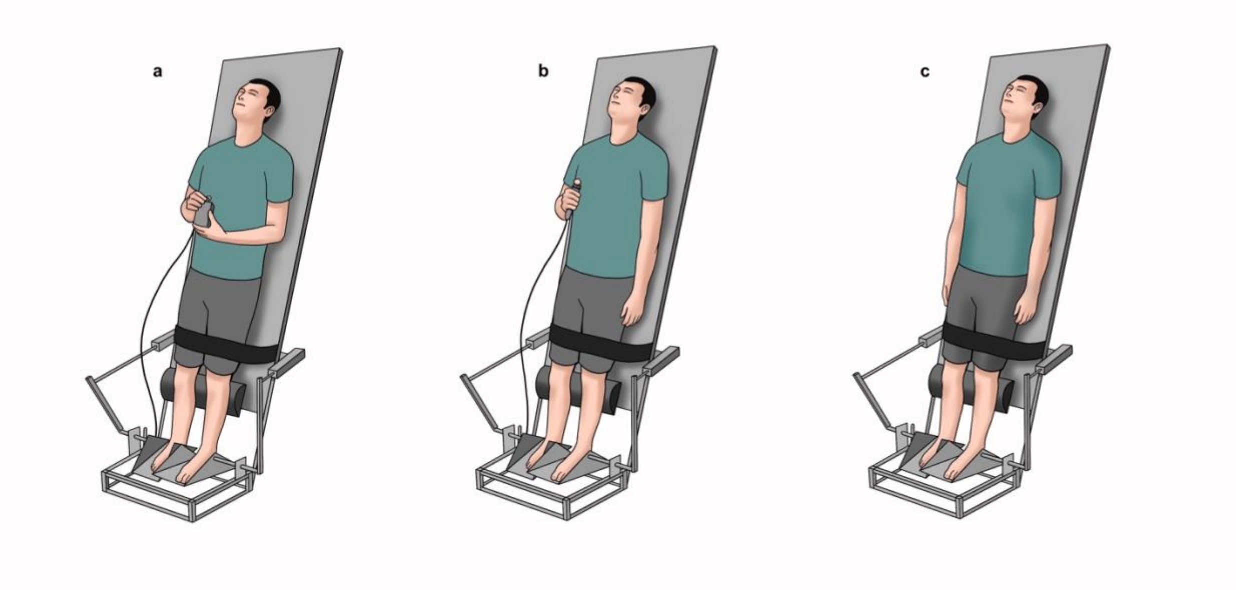


Figure 2.2: Illustrations depicting the apparatus used for all of our experiments. For the joint matching experiment subjects used a handheld dial (2.2a) to adjust the joint position (°) of the left ankle to match the target/right ankle position, while they pressed a handheld button (2.2b) for the reaction time threshold experiments, and kept their arms at their sides (2.2c) for the amplitude threshold experiments.

Participants stood barefoot while leaning against a backboard angled at 22° from vertical. Calculating the cosine of this lean angle gives a value of approximately 93%, meaning that this particular angle allowed the participant to be in a standing position with most of their body weight passing through their lower limbs (93%) while negating the need to balance. Theoretically this would allow us to measure ankle proprioception without the interference of any other factors that may affect proprioceptive acuity such as muscle tension/contraction, as it is thought that muscle spindle sensitivity is altered in actively contracted muscles. This idea is corroborated by several studies, some of which have observed that voluntary contraction appears to increase spindle sensitivity

(Behm et al., 2020) and others which have observed a decrease in spindle sensitivity following active muscle contraction (Peters et al., 2017a).

Participants were asked to close their eyes throughout the testing session so as not to receive any visual clues about foot position. Additionally, participants were asked to keep their leg muscles as relaxed as possible and to lean against the backboard. A strap was secured around their lower thighs minimising movement at the knee joint and preventing their legs from buckling while they remained in a standing position using minimal muscle activity. A foam cushion was placed behind both knees for comfort. Each foot was placed on a separate motorised footplate, each of which could be rotated independently using two linear motors attached by way of levers (Model XTA3810S; Copley Motion Systems LLC, Basildon, UK). Footplate rotation axis was made to be colinear with the ankle joint, and each footplate was equipped with encoders/sensors for position (Model CP-2UT; Midori Precisions Co., Tokyo Japan) and torque (Model 31, Sensotec Inc., Columbus, OH, USA). This positioning of participants was used for each and every one of our proprioceptive experiments.

2.4 Threshold to detection of passive motion (TTDPM)

2.4.1 Pre-piloting

Set-up and calibration of each of the footplates for position and torque were carried out first to ensure readings were as accurate and reliable as possible. Using the equation *y =*

mx + b where m = slope and b = intercept (or gain and offset respectively in practical terms), we looked for any discrepancies between actual position and torque and those that were recorded through the software. To calibrate position, we secured an inclinometer to the foot pedal with double sided tape. We then checked actual angle (reading from the inclinometer) at various angles. These measurements were then plotted against the same angles established in our MATLAB script, where the **actual** angle of 22° was set to an **analogue** angle of 0° on MATLAB, indicating the pedal was exactly perpendicular with the backboard at this point. We pressed a button while the foot pedal moved to establish points of comparison between actual angle and analogue angle and verified that the angles were identical by logging in slope and intercept where appropriate within the script. Using the same set-up, we established the ratio factor between set amplitude in MATLAB and the amplitude of footplate movement in degrees using the inclinometer. We tested a variety of amplitudes on MATLAB with a signal generator block (using a Simulink model) and then when footplate movement occurred, we went on to check the actual range of movement (maximum and minimum) recorded from the secured inclinometer. We then divided the analogue amplitude by actual amplitude to get the ratio factor and logged this into our script. This was performed for both the right and left foot pedals.

To calibrate torque, we used barbells of varying weight (0-24 kg) placing them at the centre of each foot pedal as follows:

-Right foot pedal: Centre of force was placed 20 cm from axis

-Left foot pedal: Centre of force was placed 21 cm from axis (left foot pedal was slightly longer in length)

The centre of force was outlined with a marker on each foot pedal for consistency. We went on to test the different weights, pressing a button to establish points of comparison between the right and left pedals. Keeping in mind that 1 kg = 9.81 Newtons * distance in metres (0.2 and 0.21 respectively), we added the appropriate gain and constant values to our script. Once we were certain all obtained readings were sound, we started testing our first method for TTDPM.

2.4.2 Piloting

At the beginning of our method experimentation, we aimed to determine a participants 70% detection level at each of five different rotational velocities (°/s): (0.25-0.15-0.1-0.05-.025) adopting the method by Refshauge and Fitzpatrick (1995). This method entails the participant to correctly determine the direction the footplate moved 70% of the time. For example, during a trial if we were to conduct 10 footplate movements, they would need to correctly determine at least 7 of them in order to pass that trial. MATLAB software (MathWorks Inc., Natick, MA, USA) was used to generate a motion control signal for the right footplate. This consisted of a series of 20 footplate rotations at the selected velocity, rotating either upwards or downwards in random order (with 10 up and 10 down).

At each of the above-mentioned velocities, we attempted to identify the 70% detection level by adjusting (increasing and decreasing) the amplitude of footplate displacement by 0.1° increments after each trial until the subject was able to correctly report 14 out of the 20 random movements (70%). They were asked to report:

- **“up”:** when rotation of the footplate towards dorsiflexion was perceived

  and

- **“down”:** when rotation of the footplate towards plantarflexion was perceived

An excel sheet was used to record the participants answers after each footplate motion with the following codes:

- **Up-correct (uc):** Footplate moved upwards, and participant answered correctly “up”.

- **Up-false (uf):** Footplate moved upwards, and participant answered incorrectly “down”.

- **Down-correct (dc):** Footplate moved downwards, and participant answered correctly “down”.

- **Down-false (df):** Footplate moved downwards, and participant answered incorrectly “up”.

2.4.3 The staircase method for psychophysical testing

Due to the substantial number of trials required in this first method, which proved to be extremely time consuming, we decided to explore different psychometric procedures to measure our motion detection threshold instead. Among these was the adaptive staircase method (Leek, 2001) using a transformed two-up one-down procedure, similar to that shown in Figure (2.3) from a study by Fujii and Schlaug (2013) in which the participant needed to answer two times correctly before amplitude (in this case sound (Hz)) was decreased, and amplitude was increased if they were to answer once incorrectly.

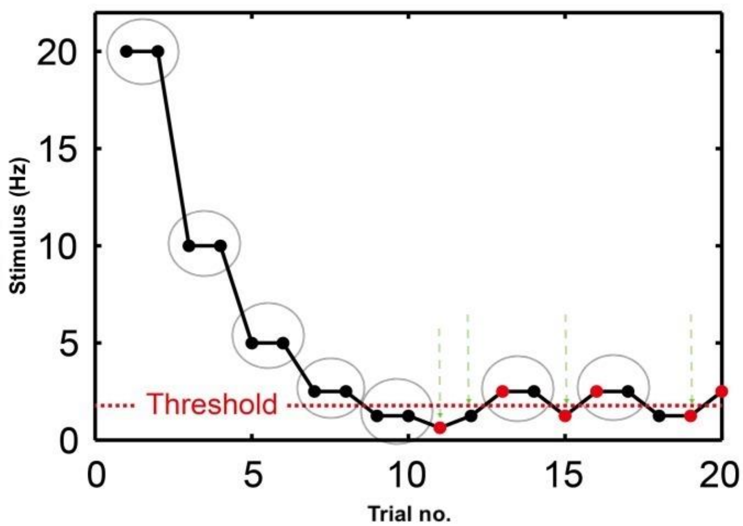


Figure 2.3: An example of a transformed staircase method, using a two-down one-up procedure. As can be seen in the figure, at the beginning of the trials the subject answers correctly twice (encircled in black) before the stimulus is decreased, “two-down”, and this keeps going on until they answer incorrectly once (indicated by the green arrows), in which case the stimulus is increased “one-up”. Threshold is then calculated by averaging the sum of reversal points, or the points at which there is a change in direction (indicated by the red dots). Adapted from Fujii and Schlaug (2013).

The idea behind the staircase method in basic terms, is to target the 50% performance level on a psychometric function that extends from 0% correct performance at chance to 100% correct performance. So, the staircase track targets the stimulus level for which the probability of a correct response equals the probability of an incorrect response, or in other words, the level at which there is equal probability the track would move up or down the stimulus spectrum. In practical terms, staircases typically begin with a high intensity stimulus, which is easy to detect. The intensity is then reduced gradually until the tested subject makes a mistake, at which point the staircase "reverses" by increasing intensity until the tested subject answers correctly, triggering another reversal. After a given number of reversals, the threshold is estimated by averaging the level of the stimulus at these reversal points in the adaptive track (Leek, 2001).

However, apart from (Geschieder, 1997), there seems to be a lack of solid guidance in the literature on the number of "reversals" needed to estimate the detection threshold, as it has been shown to vary widely among studies, from a mere handful of reversals up to several thousand (García-Pérez, 1998). According to Geschieder (1997), a total of eight reversals are usually sufficient to calculate detection threshold. Additionally, García-Pérez (1998) listed a table containing the characteristics of 77 recent studies using adaptive staircases, and it was noted that most of these studies set between 6-12 reversals to calculate detection threshold, with the majority choosing 8 reversals to terminate testing.

As to the actual testing procedure for a simple up-down staircase (Figure 2.4A), typically the track level moves after each response (one-up one-down), so there is a reduction in stimulus level when the subjects' response is positive, and likewise, there is an increase in stimulus level when the subject's response is negative. This simple form of staircase method targets the 50% performance level.

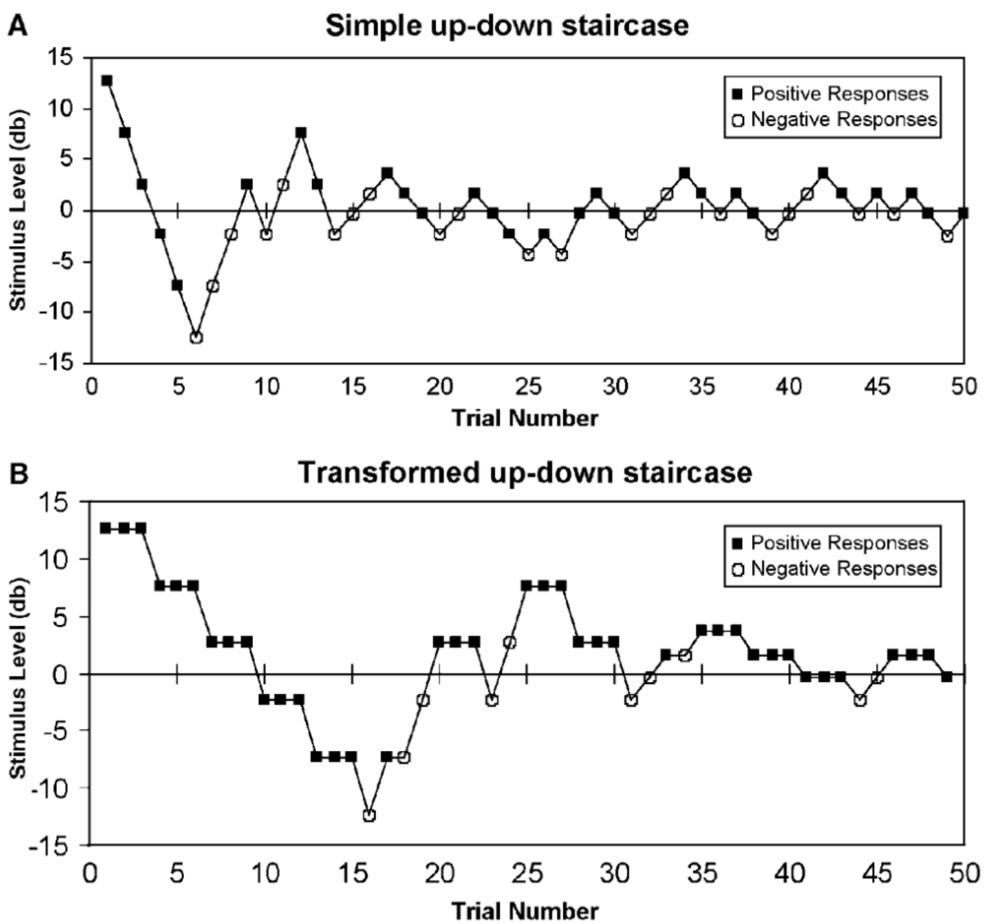


Figure 2.4: The staircase method for adaptive tracks. Following a simple (A) up-down procedure and a transformed (B) three-down, one-up algorithm (Leek, 2001).

However, to target a higher performance level, one would need to adjust to two or more positive responses for a downward movement, while the sequence for an upward movement may remain at one negative response. Recalling the method mentioned at the beginning of this section for example (two-down, one-up), in which a positive response to *two* consecutive trials is needed to move downward, and keeping in mind that the probability of a downward sequence must equal the probability of an upward sequence, we may calculate performance level as follows (Leek, 2001):

> "If p is the probability of a positive response on a given trial,
> then p*p = .5 and the target probability is $\sqrt{.5} = .707$ or 70.7%".

In the same way, a three-down, one up transformation (Figure 2.4B) leads to an even larger performance target ($p^3 = .5$, $\sqrt[3]{.5} = .794$) of approximately 80%. With the aim of improving the validity of our scores, we went on to utilise this three-down, one-up algorithm for our experiment.

2.5 Resulting methods

In this section, we demonstrate how each of our final detection threshold methods were refined for use in our experimental trials. Both methods assess movement detection threshold, although each one differs in the type of unit measured. For our first method, angular amplitude—measured in degrees (°)—was our outcome measure, whereas for our second method, reaction time—measured in seconds (s)—was the main outcome measure.

2.5.1 Method (1): Detection threshold with angular amplitude as the main outcome measure

For our new method, we needed to redesign our trial to allow for an amplitude adjustment after three sequential footplate rotations (or less) in line with our three-down, one-up algorithm. Velocity was fixed throughout the testing session (at either 0.1, 0.25, or 0.5°/s depending on which experiment was being conducted; more detail given in subsequent experimental chapters) and again movement of the footplate was on the right side only. According to Refshauge and Fitzpatrick (1995), the threshold for perceiving ankle movements is unchanged when only one ankle is moved rather than both together. This suggests that information from each ankle does not unite to facilitate perception of movement, but rather that information from each ankle is processed separately to provide information about joint movement. Therefore, testing of the right ankle seemed sufficient for our experimental needs.

During the experiment, each "step" in our staircase consisted of 3 trials or less, where the footplate moved from 0° (perpendicular to the backboard) to either upward or downward rotation for a total of three consecutive movements at most if the subject perceived all three steps correctly. If the subject were to perceive the footplate movement incorrectly once, the trials would end and the amplitude would be increased for the next "step". A random, two second "wobble" consisting of varying random degrees of dorsiflexion and plantarflexion movements separated each of the 3 consecutive footplate rotations, resetting the footplate to 0° before the next movement started. The purpose of this “wobble” was to wash out any history-dependent effects between movements (Proske et al., 1993). Figure (2.5) shows an example of a position trace using this detection method.

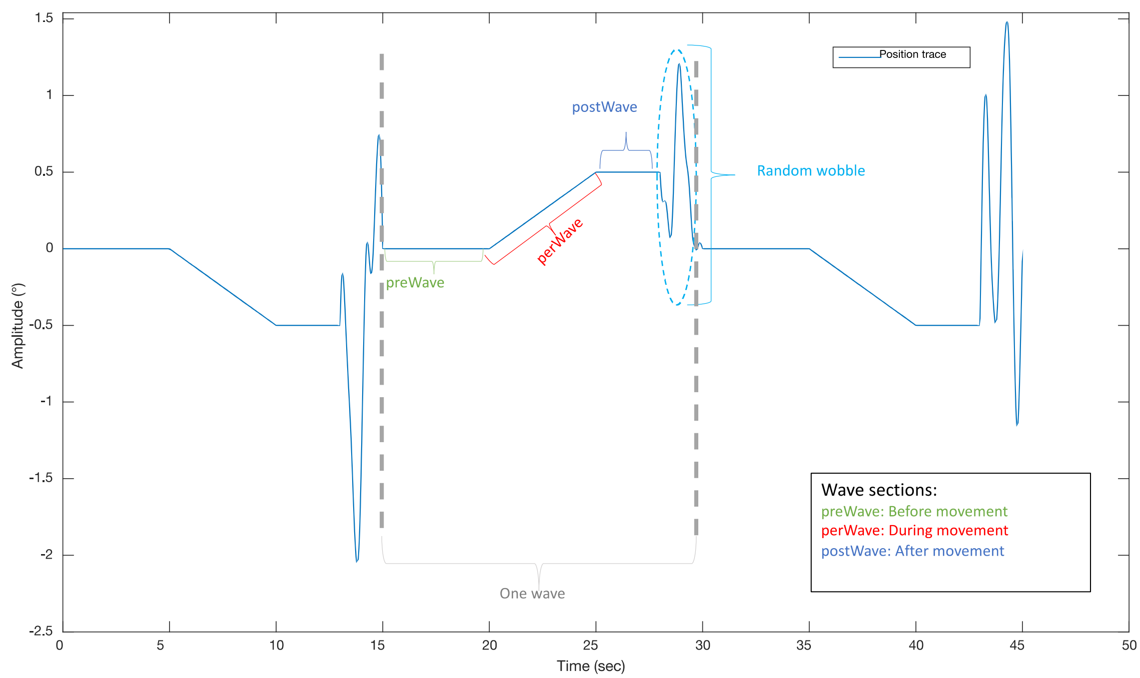


Figure 2.5: An example of a position trace taken from the right footplate during a typical testing session. The three consecutive movements here, also known as “waves”, each consist of three different parts (preWave, perWave and postWave) + a random “wobble” to reset the footplate to neutral (0°) ankle position before the beginning of the next sequential movement. In this particular example, the random footplate movements were “down”, “up”, “down”.

After some more pilot testing, we decided to incorporate this method with another form of psychomotor testing known as the “weighted staircase method”, adapted by Kaernbach (1991). In the weighted method, step *size* varies between the two different staircase directions, allowing the estimation of performance at many more target levels than what would be allowed using the transformed methods. In his paper, Kaernbach goes on to describe an example of targeting 75% correct performance with a ratio of up to down step size of 1:3 or .25/.75 where the size of a step up (Δ^+) should be 3 times the size of a step down (Δ^-). It is thought that with this weighted staircase method, there is a savings of approximately 10% experimentation time as well as a 5% decrease in error compared to traditional staircases (Kaernbach, 1991). García-Pérez (1998) also advocates the use of unequal step sizes, specifically with larger step sizes **up** due to several advantages during testing: First, they produce reversals more quickly, allowing for longer staircases with more reversals without the need for more trials. This leads to threshold estimates that are based on a more robust amount of data. Second, they almost always guarantee that reversals bracket threshold, as every reversal is interpretable as having occurred above or below threshold. Third, if after a series of lucky guesses the staircase sinks below threshold, they allow for a quick come back to a range where the stimulus is perceptible, reminding subjects what the stimulus is like. For our amplitude threshold experiment, we used our own adaptation of a combination transformed-and-weighted staircase method. We used the three-down, one-up staircase (as described above), and in line with the recommendations of García-Pérez (1998), utilised a larger step size up than down. This was achieved by *halving* the stimulus (amplitude of rotation (°) in our case) at *every other* step as follows (Figure 2.6):

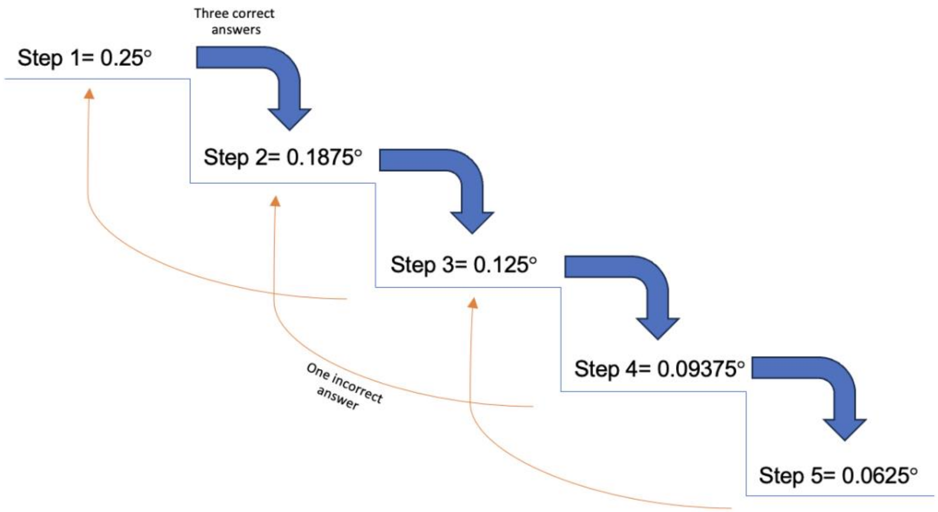


Figure 2.6: Stimulus "steps" used in our amplitude threshold experiment. Assuming we started the experiment at an amplitude of 0.25° (step 1) the subject would need to correctly perceive the direction of movement (up or down) three times at this amplitude in order to move one step down (to step 2: 0.1875°). They would again need to answer correctly three times in order to move to the next step (step 3: 0.125°) and so on. If the subject was unable to give three correct answers, the amplitude would be moved up to the step before last, so amplitude would be doubled. The amplitudes chosen for each step were such that for every *other* step the amplitude would be halved (going down) or doubled (going up).

Assuming the experiment started with an amplitude of 0.25°, the participant would need to correctly percieve the direction of movement (upwards or downwards) 3 times (at 0.25°) in order to move a step down. If they achieved 3 correct responses, the movement downward would be decreased to 0.1875° and if they were to achieve another 3 correct responses, the amplitude would be decreased to 0.125° which is half of the amplitude before last (0.25/2). However, if they were to respond incorrectly at this point we would instead increase amplitude from 0.1875 to 0.375 (so we would double the amplitude). This method of stimulus adjustment has several advantages:

First and foremost, it allows for practicality in the lab while performing the experiment live, for example when our participants surprisingly perceived footplate movement correctly further than planned, during piloting (and beyond), new smaller amplitudes needed to be calculated and logged into the software on the spot, which was an easy task due to the simple formula. Second, it is a log equation which allows for more sensitivity during testing, as both upward and downward steps get smaller and more precise as the track moves toward the lower end of stimulus intensity and likewise the steps became larger in size as the track moves toward the upper end of stimulus intensity. Using this method, we went on to pilot several participants to try and calculate their detection threshold. For all of our participants, a total of 7-8 reversals seemed sufficient to be able to calculate threshold, as steps would usually become more uniform at this point.

Figures (2.7) and (2.8) show examples of a typical session using our method for threshold detection, where velocity was fixed at 0.1°/sec.

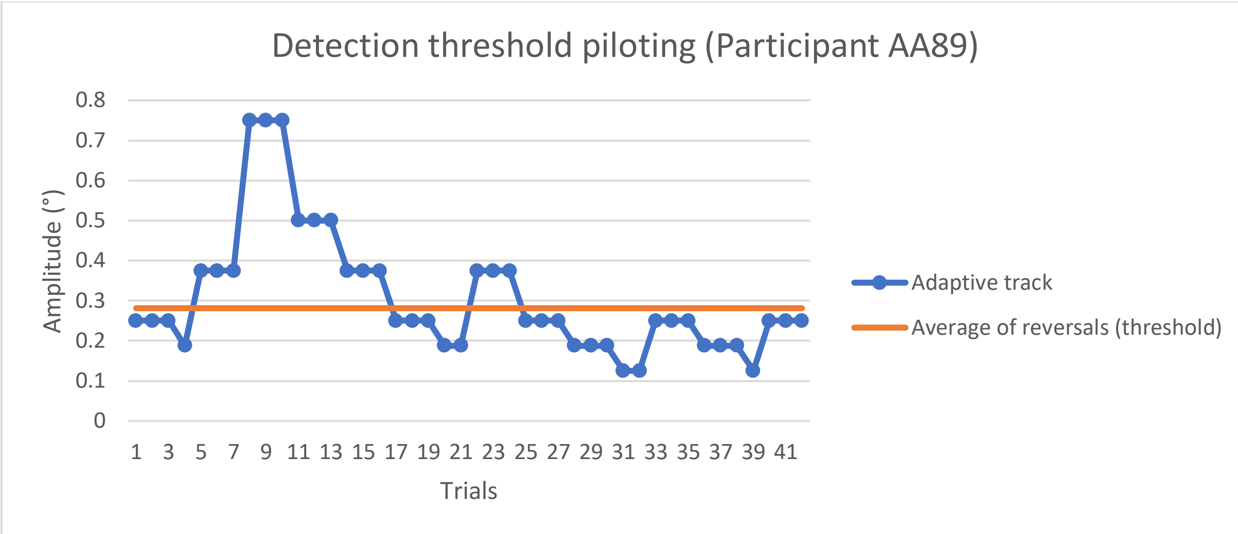


Figure 2.7: The adaptive staircase we used to obtain detection threshold for one of our participants (Example 1). In this session, it took 41 runs or trials to obtain 8 reversals (changes in direction): at .1875°, .75°, .1875°, .375°, .125°, .25°, .125°, and .25°. Therefore the average of these reversals or threshold was .28°.

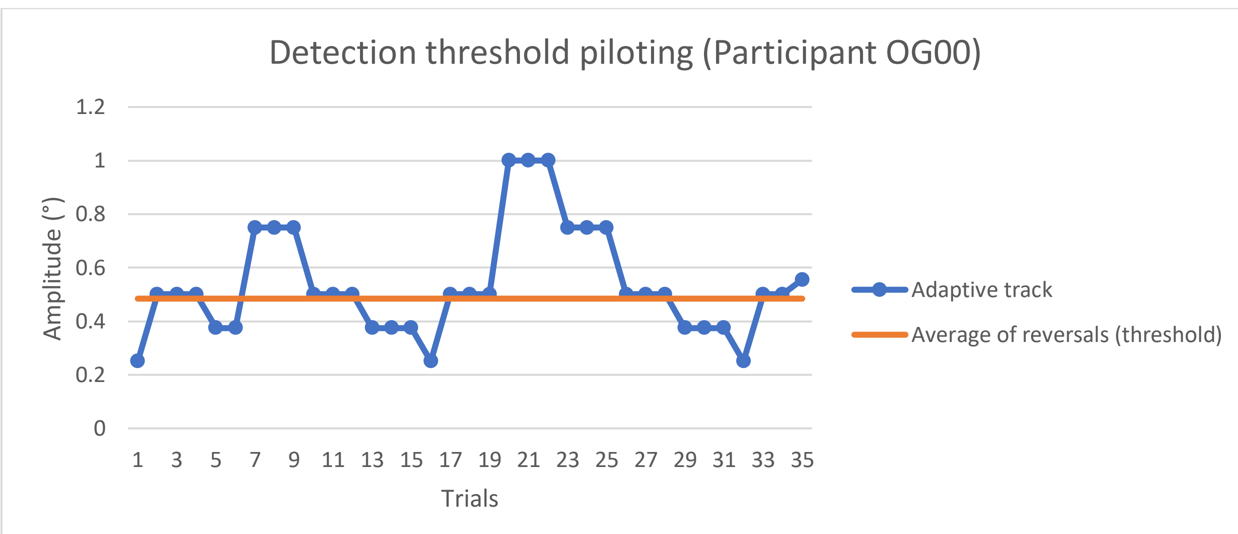


Figure 2.8: The adaptive staircase we used to obtain detection threshold for one of our participants (Example 2). In this session, it took 35 runs or trials to obtain 8 reversals (changes in direction): at .25°, .5°, .375°, .75°, .25°, 1°, .25°, and .5°. Therefore the average of these reversals or threshold was .48°.

It should be noted that a significant limitation of this method is that session time was not fixed, and was dependent upon obtaining **8 reversals**, no matter how many sessions this took. As can be seen from the examples above (Figures 2.7 and 2.8) the number of runs/trials differed between participants, and on average, with our young healthy participants, a session took approximately 35-45 minutes to complete. Although this didn't pose much of a problem at the beginning, and we went on to perform two experiments using this method, it became apparent as we used it more and more that the experimental method was very time consuming in some instances, with some participants needing upward of 58 sessions to complete 8 reversals. This would translate to over an hour of testing time in standing, and didn't seem practical as we had planned to test elderly individuals in the future.

Therefore, we decided to develop a much simpler method, with a fixed time to complete the session. Using time (s) as the outcome measure instead of angular amplitude (°), we went on to design a reaction-time threshold experiment.

2.5.2 Method (2): Detection threshold with reaction time as the main outcome measure

Using the same positioning and set-up as our previous method, we designed the MATLAB trial to instead consist of **10 consecutive rotations** moving either upwards (dorsiflexion) or downwards (plantarflexion) in random order. We tested three different velocities: 0.25, 0.5 and 0.75°/sec, and for each velocity we repeated the test three times and calculated the average reaction time for each velocity. The order of trials (based on velocity) was also randomised, so that each participant completed 9 trials, three of each velocity, in random order. During the testing session, the participant was given a handheld button and was asked to press it as soon as they detected movement, while reporting orally either “up” or “down”. Figure (2.9) shows an example positional trace of one of our reaction time testing sessions, in which the velocity was set to 0.25°/s.

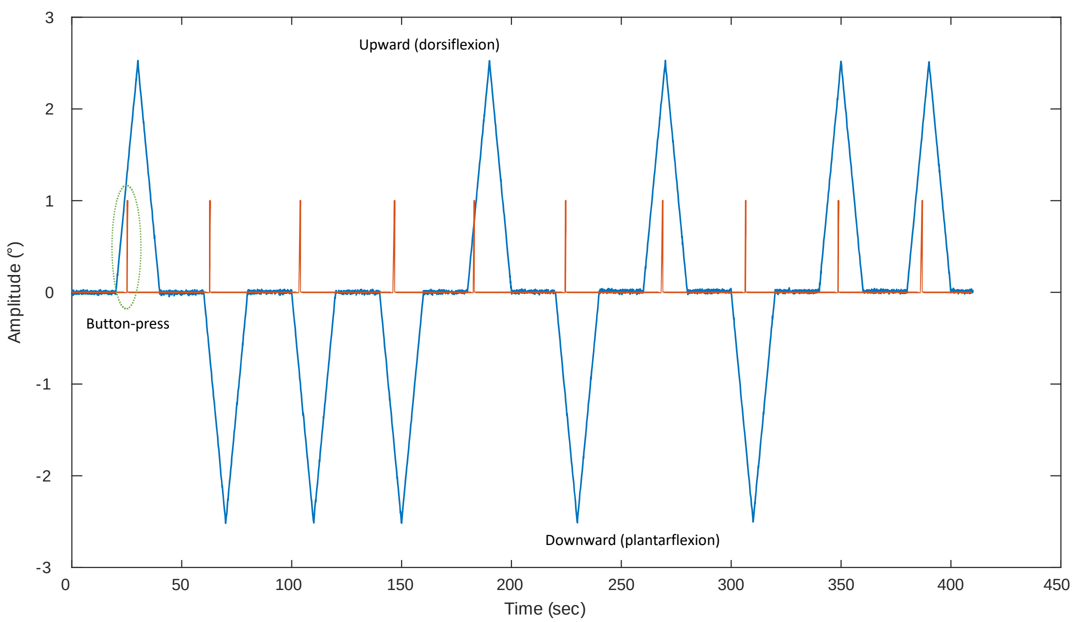


Figure 2.9: A positional trace of a pilot reaction time testing session (velocity 0.25°/s). The 10 random rotations (including 5 upwards and 5 downwards) appear in blue, with the red lines depicting the time points at which the participant pushed the button, signalling that they had detected movement. For this session, the participant detected all 10 movements correctly, and their average reaction time for this testing session was 5.8 seconds.

This new method was shown to be less time-consuming than the first, taking approximately 40 minutes to complete all trials. This was ideal as participants were required to complete a balance assessment session following this reaction time assessment, and they were able to do so comfortably, completing both legs of the experiment within an hour. As such, our reaction time method seems to be a realistic assessment tool—specifically for elderly participants—and we believe it could be an easily applicable assessment method for clinical populations with balance deficits.

2.6 Concluding remarks

Developing suitable assessment methods for our experiments was an integral part of our research, posing its own set of challenges in reaching acceptable standards for our research needs. Having said that, the process in itself proved to be an eye-opening one, requiring much critical thinking about not only the construct validity of our methods, but also about practicality during experimentation more than anything. More importantly, the resulting two methods allowed us to assess movement detection threshold in more ways than one. Our first method assesses proprioception acuity by measuring the amplitude at which a person is able to detect passive movement of their ankle joint. Our second method assesses proprioception acuity by measuring the time it takes a person to detect passive ankle movement. Both these methods are considered different versions of TTDPM, a method we found to be the one most commonly used in the literature in relation to ankle proprioception, and which has been shown to be a good predictor of standing balance in both healthy and clinical populations. As such, we believe that incorporating these TTDPM methods into our experiments will add weight to our research findings while providing additional experimental opportunities for future research in our lab.

Chapter 3: The Effect of Age and Strength upon Joint Position Reproduction

Abstract

The first of our formal experimental chapters, in chapter 3 we aimed to determine whether calf-muscle strength influenced proprioceptive acuity at the ankle joint, and whether age-related decline in proprioception is affected by muscle strength. We used a passive joint position reproduction (JPR) method to measure proprioceptive acuity at the ankle joint, and measured maximum voluntary contraction (MVC) of the calf muscle as a reference for participants' strength. A total of 36 subjects aged 20-90 years (49.3 ± 22.4) participated, and were divided into three age groups: young (20-32 years, n = 12) , middle-aged (33-68 years, n = 12) and old (69-90 years, n = 12) . Our two main outcome measures were absolute error (AE)(°) and leg strength (N.m). Our secondary outcome measures were constant error (CE)(°), variable error (VE)(°), in addition to joint torque and joint torque variability (N.m). Results showed a significant positive correlation between age and mean AE specifically at neutral angles of the ankle joint (0° and 2°)($r_{(34)} = 0.33$, $p = 0.046$, $r_{(34)} = 0.42$, $p = 0.01$ respectively). We also found a significant negative correlation between leg strength and AE while the ankle was rotated slightly towards dorsiflexion (4° and 6°)($r_{(34)} = -0.37$, $p = 0.028$, $r_{(34)} = -0.34$, $p = 0.044$ respectively). Furthermore, when analysing our data based on age groups, we found that higher strength significantly predicted AE for the middle-aged group ($t_{(10)} = 2.52$, $p = 0.03$), but not for the young or old age groups. In conclusion, our findings suggest that although age significantly

predicted proprioceptive acuity through JPR at the ankle joint, matching error did not improve as a result of higher muscle strength except for the middle-aged participants.

3.1 Introduction

Ankle proprioception plays a key role in our ability to maintain adequate standing balance through sensory input from the muscles surrounding the ankle. This input has been shown to be more sensitive than both visual and vestibular input in perceiving body sway (Fitzpatrick and McCloskey, 1994), making it an ideal sensory modality to study in the context of balance, and also its potential role in susceptibility to injurious falls due to old age.

3.1.1 Literature review

The effect of ageing on proprioceptive sensation has been a topic of debate in recent years, with conflicting results challenging the once well-accepted concept that proprioceptive acuity, as with other senses in the body, deteriorates as we age (Ribeiro and Oliveira, 2011). Ankle joint position sense in particular, which is typically assessed using the joint position reproduction (JPR) technique, has been shown to be negatively affected by ageing (Ribeiro and Oliveira, 2007). Other studies showing a significant age-related decline in ankle proprioception include Verschueren et al. (2002), Deshpande et al. (2003) and Westlake et al. (2007). Verschueren et al. (2002) observed this decline using a method other than JPR for assessing proprioception: the threshold to detect passive movement (TTDPM) technique. Whereas Deshpande et al. (2003) and Westlake et al. (2007) used JPR alongside TTDPM in their studies.

Conversely, a robust study by Djajadikarta et al. (2020) found a negligible effect of age on ankle proprioception performance in healthy participants when movement history of the ankle was controlled. Previous history of muscle contraction and length changes has been shown to influence proprioception, specifically position and movement sense in the passive limb (Proske and Gandevia, 2012). Therefore, controlling for movement history may decrease potential errors related to the thixotropic behaviour of passive muscles. The study incorporated three proprioceptive testing methods (TTDPM, proprioceptive reaction time, and JPR), and participants were placed in a seated testing position with knees extended and ankles dorsiflexed, a posture shown to be associated with improved proprioceptive performance at the ankle (Refshauge and Fitzpatrick, 1995). Results suggested that there is little effect of age on ankle proprioceptive performance in healthy community-dwelling people. Although these results are surprising, it should be noted that while evidence that muscle spindles—the mechanoreceptors arguably most important for proprioceptive sensation—undergo age-related anatomical and physiological changes at the peripheral level has been provided, observations of functional outcomes remain limited (Proske and Gandevia, 2012).

Nonetheless, research has shown that there is an association between sarcopenia (the loss of muscle mass that occurs during normal ageing) and the loss of motor units, with intramuscular electromyography (EMG) reporting 30% to 50% fewer motor units in human subjects by the age of 70 years (Narici and Maffulli, 2010; Piasecki et al., 2019). This loss has been shown to be more pronounced in the lower limbs, reducing *muscle power* by as much as 60% in the elderly (Proske and Gandevia, 2012). Also, it has been reported that the highest strength decline related to ageing is that of the plantar flexor muscles (Simoneau et al., 2005).

However, although older muscle contains fewer motor units, the remaining motor units are approximately 50% enlarged and are reported to show increased complexity and fibre density. This is said to be a compensatory mechanism in response to the loss of motor units (Piasecki et al., 2016). It is believed that regular exercise is associated with the improved fibre reinnervation found in older adults which would lead to increased motor unit *size* thus slowing age-related loss of muscle mass (Proske and Gandevia, 2012; Mosole et al., 2014).

Moreover, numerous studies indicate the beneficial effects of exercise and physical activity on balance (Lord et al., 1993; Mahmoud et al., 2014) and lower limb proprioception (Tsang and Hui-Chan, 2003; Ribeiro and Oliveira, 2010; Wang et al., 2016; Niespodziński et al., 2018). Ribeiro and Oliveira (2010) found that among the four groups they studied (exercised-old, n = 31; non-exercised-old, n = 38; exercised-young, n

= 35; non-exercised-young, n = 25) joint position sense was not only superior in the exercised old age group compared to the non-exercised old age group, but it was similar to that of the non-exercised young age group (all $p<0.05$). This suggests that exercise may counterbalance the age-related decline in proprioceptive acuity often seen in older subjects.

Furthermore, in a study by Wang et al. (2016), a significant relationship was not only found between joint position sense and both static ($p< .05$) and dynamic balance ($p< .01$), but also between static and dynamic balance and quadriceps strength ($p< .05$). To add to that, Guney et al. (2016) found a direct correlation between quadriceps strength and joint position sense ($p< .05$). In addition, Butler et al. (2008) found that subjects with weaker dorsiflexors exhibited greater postural sway than those with strong dorsiflexors, but only when they were forced to rely upon proprioception by closing their eyes. Taken together, these studies indicate that muscle strength may have impact on proprioceptive acuity to some extent.

For older aged adults, particularly those susceptible to falls, finding an evidence-based strategy to maintain ankle proprioception sensitivity would be paramount in their ability to perform daily tasks safely and independently.

3.1.2 Aim of study

The current study aimed to examine whether muscle strength influenced proprioceptive acuity, and specifically, to determine whether age-related decline in proprioception is related to changes in muscle strength.

We analysed the relationship between age, proprioceptive acuity, and muscle strength, to establish whether muscle strength is associated with proprioceptive acuity, particularly among older individuals.

To do this, proprioception was evaluated through JPR, which is a commonly used method to assess proprioception. It involves measuring the accuracy of replicating joint position in the absence of visual input. Proprioception was measured at the ankle joint and calf muscle strength was also assessed, with the maximum voluntary contraction (MVC) recorded and used as the reference for a person's strength.

3.1.3 Hypotheses

For our study, we hypothesised that age influences proprioceptive ability, and that this influence will be related to changes in muscle strength.

Null Hypotheses (H_0):

-Age will have no effect on ankle proprioception.

-Changes in muscle strength will have no effect on proprioceptive acuity at the ankle.

Alternative Hypotheses (H_a):

Age will have a direct effect on ankle proprioception and this effect will be related to changes in muscle strength among individuals.

The current study will determine whether a relationship does indeed exist between contractile and sensory muscular processes during a proprioceptive task, and whether a compensatory mechanism between these processes may come into play for stronger older-aged individuals.

3.2 Methods

3.2.1 Participants

A total of 36 healthy participants (15 females) aged between 20-90 years, with a mean age of 49.3 ± 22.43 years, an average height of 172.8 ± 8.67 cm, and an average BMI of 23.9 ± 3.23 kg/m^2 were included in the study. Exclusion criteria for the experiment included any ongoing physical or neurological injury or illness which might affect normal control of standing balance and/or any injury to the lower limbs. The experiment took place in the Human Balance Laboratory in the School of Sports, Exercise and Rehabilitation in the University of Birmingham, and took approximately one hour to complete per participant. The experiment was approved by the School of Sport, Exercise and Rehabilitation Ethical Review Committee at the University of Birmingham (ERN_14-0740A) and performed in accordance with the Declaration of Helsinki (World Medical Association, 2013). Participants were given information sheets on the experiments and provided written, informed consent before starting the trials.

3.2.2 Experimental set-up

Testing was performed with the participant standing barefoot with their back leaning against a backboard angled 21.8° from vertical, with a Velcro strap fastened around the lower thighs to minimise any movement of the knees (Figure 3.1).

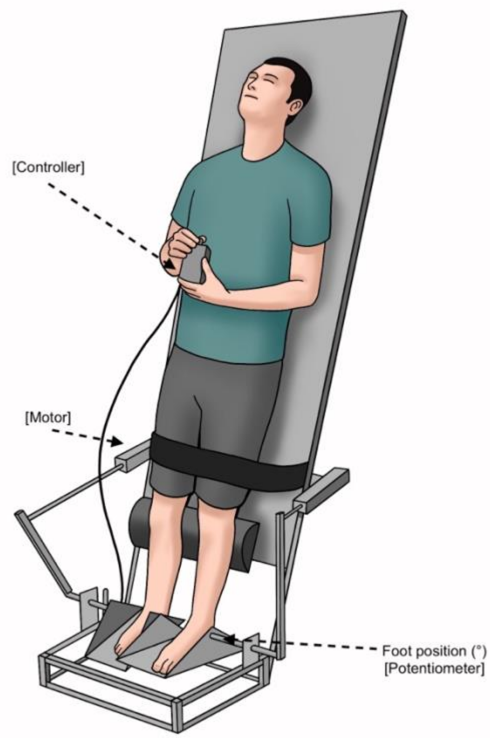


Figure 3.1: Participant positioning while performing the matching task. The board was angled at approximately 22° from the vertical, and the footplates were positioned perpendicular to the backboard (considered here to be an ankle angle of 0° or "neutral") at the start of each trial.

A foam cushion was placed behind both knees for comfort. This setup was chosen to ensure the person was bilaterally weight bearing without the need to balance themselves as they would when standing normally. It also allowed them to maintain a standing position using minimal muscle activity without collapsing. Each foot was placed on a separate motorised footplate, which could be rotated individually using two linear motors attached by levers (Model XTA3810S; Copey Motion Systems LLC, Basildon, UK). The axis of rotation of each footplate was made to be approximately collinear with the ankle joint. In addition, each foot plate was affixed with sensors for position (Model CP-2UT; Midori Precisions Co., Tokyo, Japan) and torque (Model 31, Sensotec Inc., Columbus, OH, USA). Participants were asked to remain relaxed through their legs, not pressing with any force on the footplates apart from their own body weight.

3.2.3 Procedure

The experiment consisted of a matching task in which the participants' right foot was passively moved into a random pre-determined target angle, any one of 7 target angles (-6°, -4°, -2°, 0°, 2°, 4°, 6°) where 0° refers to the neutral position in which the footplate is perpendicular to the backboard. The participant was given a hand-held dial which controlled the left foot motor. This allowed them to manually reposition their left foot to match the right foot position as accurately as they could perceive with their eyes closed, after which they pressed a button to submit their answer/position. After the button was pressed, a random two-second wobble —which was generated by the sum of five sinewaves with frequencies that randomly varied from 0.5 to 1.5 Hz—was initiated at the right footplate before it rested at its new position. The purpose of this wobble was to minimise any short-term memory-dependant and/or thixotropic effects the previous foot position may exert upon joint position sense during the following trial. Each pre-determined target angle was included within a block of seven trials, and a total of ten blocks were tested resulting in 70 trials per participant. Within each block, the order of target angles was randomised. Participants were allowed short periods of rest between trials and were not provided with any performance feedback throughout the testing session. However, there were no time constraints imposed for the experiment which meant participants could perform the matching task in a self-paced manner.

To assess calf muscle strength, an isometric maximum voluntary contraction (MVC) force (N.m) was measured for each participant immediately following the testing session. Participants were seated with their knees flexed and their thighs clamped from above so that the knee was flexed approximately 85° from the horizontal. Arms were crossed in front of their chests, and they were asked to push upwards against a strain gauge using their plantar flexors. They received visual feedback of their force output on a computer screen, and they repeated this plantar flexor force several times with brief 30 second rest periods in between. Maximum force was recorded when the peak force plateaued, and this was recorded as their MVC value.

Position and torque measurements obtained from the footplates, in addition to the strain-gauge signals were sampled at 100 Hz using Simulink Real-Time Desktop (MathWorks Inc., Natick, MA, USA)

3.3 Analysis

GraphPad Prism (GraphPad Prism, GraphPad Software, Boston, Massachusetts USA) and SPSS software (IBM SPSS Statistics, IBM Corp, Armonk, NY, USA) were used for data analysis. Data was analysed in two ways: First by performing analyses on the group, then by separating participants by age group into three subgroups:

-Young (n = 12, 20-32 years): Mean age of 23.5 ± 2.94 years, average height of 175.8 ± 7.52 cm and average BMI of 23.01 ± 1.93 kg/m^2.

-Middle-aged (n = 12, 33-68 years): Mean age of 49.9 ± 11.84 years, average height of 175.9 ± 8.96 cm and average BMI of 24.54 ± 4.62 kg/m^2.

-Old (n = 12, 69-90 years): Mean age of 74.5 ± 5.65 years, average height of 166.6 ± 6.3 cm and average BMI of 24.27 ± 2.59 kg/m^2.

For analyses purposes, **Foot Position** is referred to the angular position of the left footplate at the point when the participant pressed the button to indicate they perceived the left footplate position to be “matched” to the right. For each of the 7 target positions, the average matched position was calculated from 10 trials giving us 7 average matched positions for each participant.

Three variables: Absolute Error (AE), Constant Error (CE) and Variable Error (VE) were used to assess joint matching performance (Reynolds, 2005; Reynolds et al., 2020). For our experiment, AE is referred to the absolute difference between left and right footplate position, which demonstrates overall performance. CE refers to the signed (+, -) difference between left and right footplate positions, reflecting performance bias. VE is defined as the standard deviation of left footplate position for each target angle and reflects precision of matching performance.

A fourth variable, Joint Torque (JT), which refers to the passive resistance of the left ankle joint against the footplate, was also used in our analyses. This variable was obtained from a Simulink signal trace recorded for each of our participants during the JPR trials.

Data was tested for normality using the Shapiro-Wilk test, and was found to be normally distributed ($p>0.05$). As such, parametric tests were selected for our inferential analyses. After performing descriptive analyses on our data, we used a repeated measures analysis of variance (ANOVA)—with a Geisser-Greenhouse correction applied to the degrees of freedom—to look for significant differences in AE, CE and VE between target angles for group data.

As participants were also divided into three age groups: young, middle-aged and old, a one-way ANOVA was performed between these age groups to explore the potential effects of age on the data, and correlation analyses were used to investigate any effects of age and strength on AE and torque.

Additionally, participants were grouped based on strength (lower strength/higher strength) to explore the effect of strength on AE and torque further.

Statistical significance was set to $p < 0.05$. All results in the main text are reported as mean ± standard deviation.

3.4 Results

3.4.1 Demographic data

A total of 36 subjects (21 males, 15 females) aged between 20-90 years (mean of 49.3 ± 22.4 years) with an average BMI (23.94 ± 3.2 kg/m^2) participated in the study.

3.4.2 Descriptive and inferential data

Results of our analyses will be presented under the following subcategories:

1. Foot position
2. Absolute error
    a. Group analyses
    b. Age group analyses
    c. Strength and AE

3. Constant error
    a. Group analyses
    b. Age group analyses
4. Variable error
    a. Group analyses
    b. Age group analyses
5. Joint torque
    a. Group analyses
    b. Age group analyses
6. Strength

1. Foot Position:

Figure (3.2) shows left (matched) foot position (°) data of a single participant compared to target foot position (dotted line). Undershooting of the matched foot position is clear here for all target foot positions, which was also reflected in group data, presented as average left (matched) foot position (Table 3.1) and individual matched foot position for each of the 36 participants (Figure 3.3). Foot position here represents matching accuracy, which appears to be greatest at angles 0° and 2° (Table 3.1) and (Figure 3.3) decreasing inversely as the angle increased both ways with the lowest accuracy at 6° and -6°.

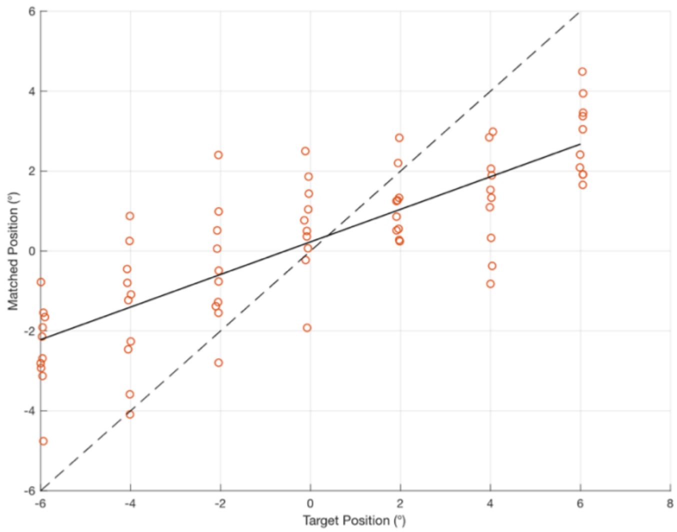


Figure 3.2 (left): Ankle matching for a single participant. Right and left feet were the target and matched positions, respectively. Clear undershooting of the average matched position (solid line) can be observed here, which is also reflected in group average left foot position (Figure 3.3).

Target angle	-6°	-4°	-2°	0°	2°	4°	6°
Mean (°)	-2.897	-1.821	-0.627	0.2593	1.372	2.197	3.584
Standard Deviation	1.802	1.474	1.420	1.283	1.159	1.278	1.578

Table 3.1: Descriptive statistics for left foot position (matched) for each target angle.

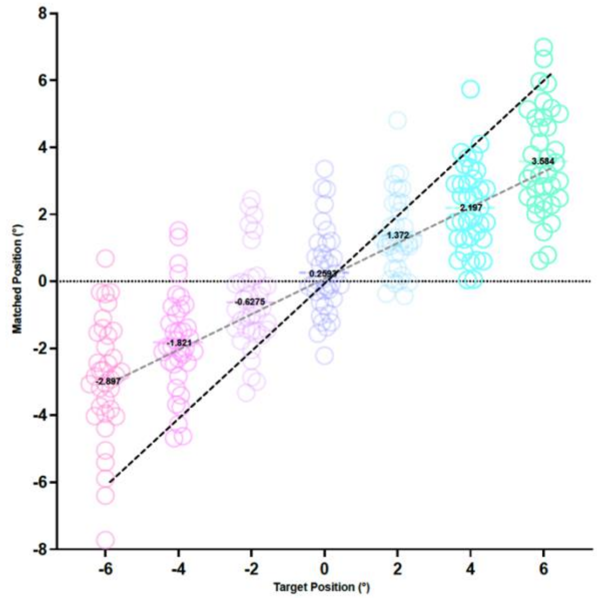


Figure 3.3 (right): Group ankle matching data including average matched foot position. Accuracy was best at 0° and 2°, proportionally decreasing as the target angle increased. Worst accuracy was found at the largest angles (6° and -6°) with matched position reaching only approximately 50% of target position.

There is a clear undershooting of target position for all angles; apart from 0°. The undershoot is particularly obvious for negative angles. Pairwise comparisons (corrected for multiple comparisons using the Tukey test) revealed a significant effect of target angle on left foot position between each angle pair (*****p < 0.0001*).

A repeated-measures analysis of variance (ANOVA) performed on group data revealed a significant effect of target angle on left (matched) foot position ($F_{(1.292, 45.23)} = 196.2$, *p < 0.0001*). In addition, post hoc analysis revealed this significance existed between every angle pair (*p<0.05* for all comparisons)(Figure 3.3). This confirms that participants were effectively able to differentiate between each of the target angles. In addition, 85% of variation in matched position could be explained by target position ($R^2 = 0.85$).

2. Absolute Error (AE):

a. Group analyses

Matching error was calculated as the absolute difference between target and matched foot position for each angle. Individual AE scores can be seen in Figure 3.4.

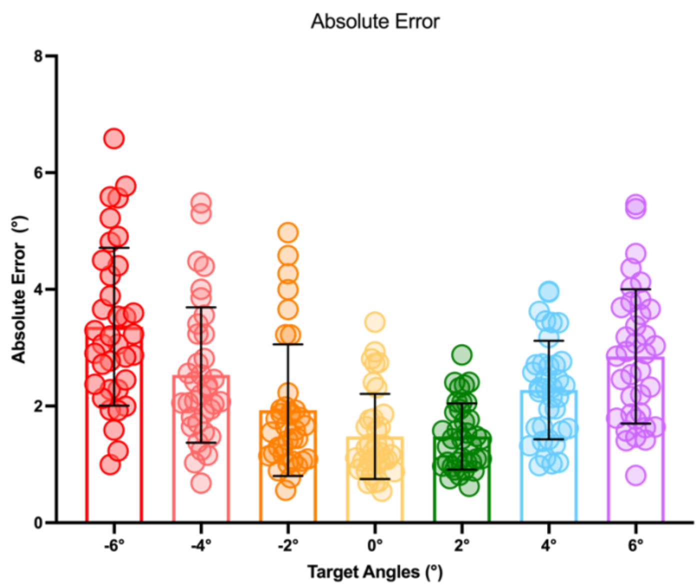


Figure 3.4: Absolute error at each target angle. Individual responses are shown, alongside group mean and standard deviation bars. Matching was best at 0° and 2° while it was least variable at 2°.

Target angle	-6°	-4°	-2°	0°	2°	4°	6°
Mean (°)	3.359	2.532	1.929	1.479	1.479	2.275	2.850
Standard Deviation	1.354	1.161	1.127	0.7309	0.5666	0.8448	1.152

Table 3.2: Descriptive statistics for absolute error for each target angle.

A repeated measures ANOVA indicated a significant effect of target angle on AE scores ($F_{(1.768,\ 61.89)} = 21.93$, $p < 0.0001$). Mean AE was lowest at 0° and 2°, increasing proportionally with the increase in angle both ways (Table 3.2). In addition, AE standard deviation was lowest at 2°, 0°, and 4° (in that order-Table 3.1) indicating that data was least variable at these angles. Together these results indicate that participants tended to have the most accurate and precise matching ability when the target foot position was neutral or slightly dorsiflexed.

Correlation analyses revealed a significant negative correlation between age and leg strength ($r_{(34)} = -0.6$, $p < 0.0001$) and a significant positive correlation between age and mean AE for all target angles ($r_{(34)} = 0.45$, $p = 0.0065$) (Figures 3.5.1 & 3.5.2 respectively). This would indicate that older participants tend to be worse in general at matching than younger participants. However, upon exploring individual angles, it was shown that age had a significant positive effect on AE scores specifically at angles **0°** and **2°** ($r_{(34)} = 0.33$, $p = 0.046$, $r_{(34)} = 0.42$, $p = 0.01$ respectively), while leg strength had a significant inverse effect on AE scores at angles **4°** and **6°** ($r_{(34)} = -0.37$, $p= 0.028$, $r_{(34)} = -0.34$, $p = 0.044$ respectively).

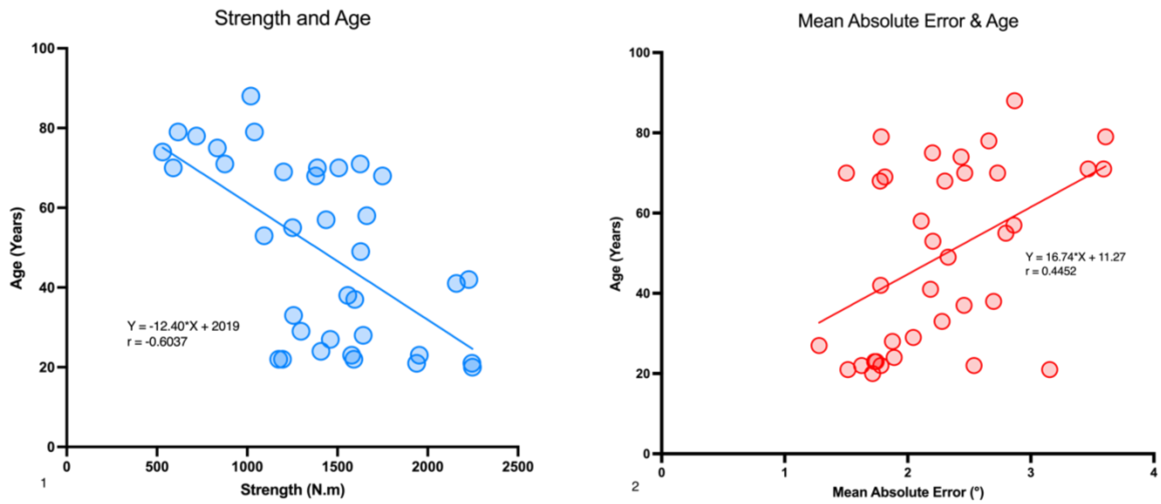


Figure 3.5: Effect of age on strength and mean AE. (1) A moderate negative correlation was found between strength and age (*p < .0001),* and (2) a moderate positive correlation between AE and Age (*p =0.0065).*

b. <u>AE within age groups</u>:

Furthermore, data was analysed to check for any trends within each age group. It was found that this data followed the same pattern as table (3.2): Absolute error was shown to be lowest at 0° and 2° within each age group, increasing proportionally with increase in angle both ways. Additionally, a one-way ANOVA confirmed that, consistent with whole group data, there was a significant effect of target angle on AE in young ($F_{(6, 77)}$ = 4.82, $p = 0.0003$), middle-aged ($F_{(6, 77)}$ = 15.23, $p < 0.0001$), and old age groups ($F_{(6, 77)}$ = 3.498, $p =0.004$) separately. A comparison of mean and standard deviation of AE for each age group can be found in Appendix (1).

c. AE and Strength within age groups:

Looking at the effect of strength on AE within each age group, it appears only the young age group exhibited a significant effect of strength at angles **6°** ($r_{(10)}$ = -.74 , *p =0 .006*) and **-2°** ($r_{(10)}$ = .68, *p = 0.015*). No effect of strength was found in the other two age groups.

3. Constant Error (CE):

a. Group analyses:

In our study, CE represents the difference between left and right footplate positions including their direction (+, -) reflecting performance bias. As expected, CE data showed positive values for target angles -6°, -4° and -2°, and negative values for target angles 2°, 4° and 6° indicating that participants' left "matched" foot never quite reached the position of the right foot (so matching was underestimated) for all target values except for 0°. A table with mean CE values at each target angle can be found in Appendix (2).

b. <u>CE within age groups</u>:

Looking at CE in terms of age group (Figure 3.6), it's apparent that young participants were closest to target position for all positions except 0° which was closest among the middle-aged group. However, a one-way ANOVA showed no significant difference between age groups in terms of CE ($F_{(2,33)}$=0.42, *p=0.66*).

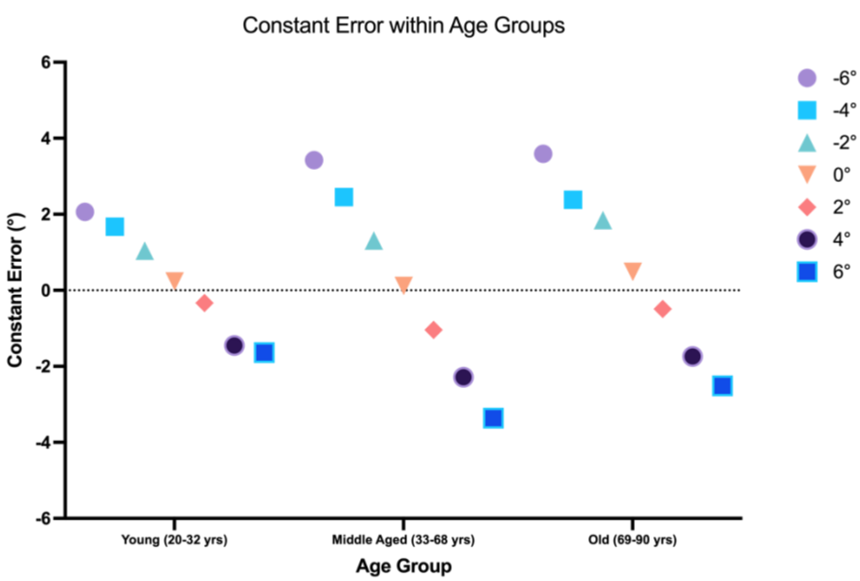


Figure 3.6: Comparison of average CE scores within each age group. Data shows a consistent underestimation of matched foot position, with the young age group showing the least deviation from target position overall. Symbols depict target angles (See legend).

4. Variable Error (VE):

a. Group analyses:

As mentioned previously under the "foot position" section, matched position *accuracy* appeared to be greatest at neutral angles 0° and 2°. Similarly, matched position *precision* (represented by average VE scores) was best among participants at angle 2° (Figure 3.7). A repeated measures ANOVA indicated an effect of target angle on variable error ($F_{(3.443, 120.5)} = 2.654$, *p =0.04*) however post hoc analysis revealed the significance to be between angles 0° and 2° only (*p=0.035*) (Figure 3.7).

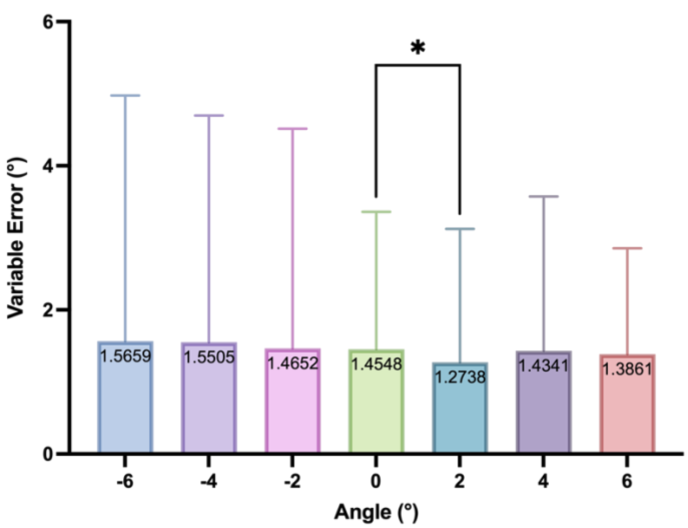


Figure 3.7: Mean VE for each target angle. VE was lowest at angle 2° indicating participants had the highest precision at this target angle. A repeated-measures ANOVA indicated an effect of target angle on variable error (p = .04). Post hoc analysis (corrected for multiple comparisons using the Tukey test) indicated the significance to be between angles 0° and 2° only (p=0.35).

b. VE within age groups:

Figure 3.8 illustrates the spread of foot position data points (which is another way to visualise VE) within *and* between subjects of each age group. Figure 3.9 shows VE data for each age group. Although the middle-aged group showed the least variation in foot position for all target angles (Table 3.3), a one-way ANOVA found no significant difference between age group VE ($F_{(2,33)}$=2.241, $p=0.122$).

A table containing mean and standard deviation values for foot position based on age group can be found in Appendix (3).

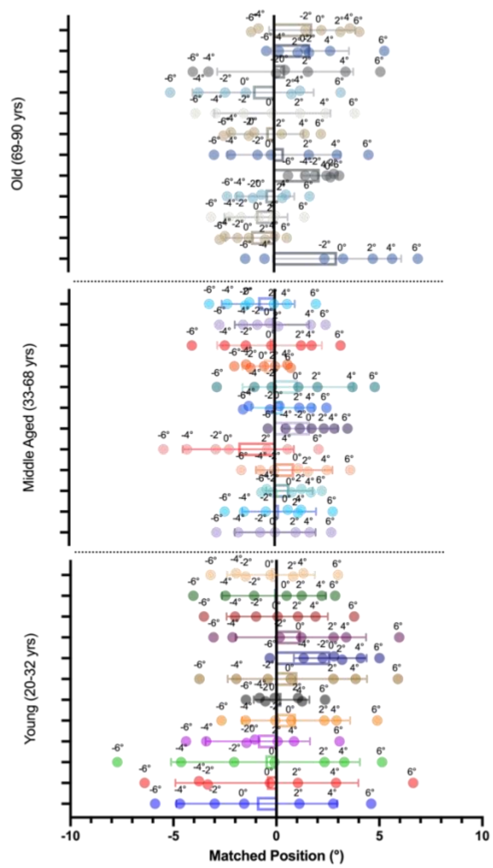
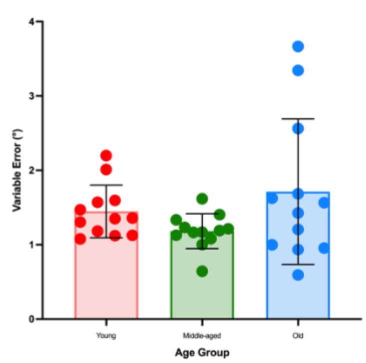


Figure 3.8 (left): A comparison of left (matched) foot position data spread. Comparisons may be viewed within and between subjects of the young, middle-aged, and old age groups. Each "matched" point is labelled with the target angle.

Figure 3.9 (below): A comparison of VE among age groups (mean and standard deviation). Although lowest for the middle-aged group,



Target angle	-6°	-4°	-2°	0°	2°	4°	6°
Young (20-32 yrs.)	1.53	1.49	1.39	1.35	1.24	1.56	1.56
Middle-Aged (33-68 yrs.)	1.20	1.27	1.18	1.29	0.98	1.22	1.14
Old (69-90 yrs.)	1.96	1.89	1.83	1.73	1.60	1.52	1.46

Table 3.3: Mean variable error (VE) for each target angle among age groups. The middle-aged group had the lowest VE among age groups for all target angles. However, a one-way ANOVA found the difference to be non-significant (p>0.05).

5. **Joint Torque (JT):**

a. <u>Group analyses</u>:

During the experiment, participants were asked to keep their legs completely relaxed, without pressing with any force on the foot plates apart from their own body weight. It was crucial that participants did not contract their lower leg muscles to eliminate any effect muscle contraction may have on proprioceptive acuity at the ankle. There will however remain a passive JT force, which refers to the passive resistance of the left ankle joint against the footplate, which corresponds to the amount of tension that exists in the calf muscle. It is useful to record JT as it may have implications on proprioceptive ability throughout the ankle joint range of motion tested.

An expected significant correlation was found between BMI and left torque as well as BMI and right torque at all target angles ($p < 0.05$).

Additionally, a repeated measures ANOVA indicated a significant effect of target angle on left torque ($F_{(1.385, 48.47)} = 109.9$, $p < 0.0001$) with $R^2 = 0.76$ as well as right torque ($F_{(1.346, 47.10)} = 316.8$, $p < 0.0001$) with $R^2 = 0.9$. This demonstrates that 76% of left torque and 90% of right torque variance could be explained by target angle, which, taken with our descriptive torque results (Appendix 4) indicates that torque tended to increase proportionally with target angle (towards dorsiflexion), with a slightly stronger tendency on the right side.

To confirm participants exhibited similar force in left and right lower limbs during trials, torque was compared between both sides. A paired t-test confirmed there was no significant difference between right and left side torque ($t_{(35)} = 0.03$, $p = 0.98$). Therefore, it may be safe to assume that participants were indeed relaxed during testing, and that matching ability was indicative of sensory receptor (muscle spindle) influence.

b. JT within age groups:

Although the middle-aged group had the highest average BMI among the age groups, a one-way ANOVA indicated there was no significant difference between age groups' BMI ($F_{(2,30)}$=1.1, *p=0.346*). Additionally, a one-way ANOVA confirmed that the difference in left and right torque between young, middle-aged and old age groups was not significant ($F_{(2,33)}$=0.59, *p=0.56* and $F_{(2,33)}$=1.94, *p=0.16* respectively).

6. Strength among age groups:

To explore the effect of strength further, data was analysed within each age group to account for the effect of age on matching ability. The same three age groups (young, middle-aged and old) were each split into two groups: **Higher strength** and **lower strength**. Figure 3.10 compares mean AE between higher strength and lower strength participants within each age group.

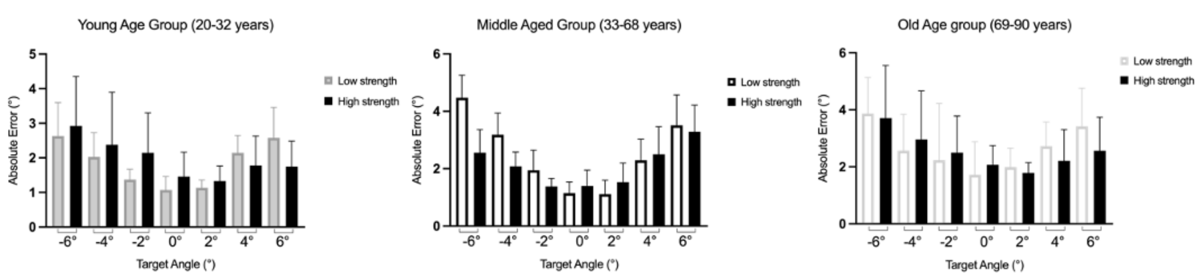


Figure 3.10: A comparison of mean AE scores between low and high strength groups. Data of young, middle-aged and old age groups may be compared based on strength.

Although AE scores increased (in general) the farther away from neutral they became, lower strength individuals tended to show better matching ability (lower AE scores) at neutral angles, whereas higher strength individuals tended to show better matching ability towards the end ranges, specifically in the direction of dorsiflexion. This applied to all three age groups. Furthermore, average AE (all target angles) appeared *lower* for the higher strength groups in both the middle-aged and old age groups, however an unpaired t-test revealed this difference was only significant for the middle-aged group ($t_{(10)} = 2.52$, $p = 0.03$). Descriptive data comparing the difference in AE between higher strength and lower strength individuals for each age group can be seen in Appendix (5).

Torque values between higher strength and lower strength participants were also compared for additional clarity on the previous findings. Higher strength groups tended to show lower mean torque than lower strength groups. However, an unpaired t-tests showed there was no significant difference between the two. A table comparing left JT values for lower strength and higher strength groups among age groups can be viewed in Appendix (6).

3.5 Discussion

The current study aimed to assess the effect of age and strength on proprioceptive acuity at the ankle joint. Joint position reproduction (JPR) was the chosen method for proprioceptive testing, with matching performed passively on the contralateral (left) ankle.

3.5.1 Foot position and AE

Participants were able to significantly differentiate between the 7 target angles (-6°, -4°, -2°, 0°, 2°, 4°, 6°) during testing. This indicates that participants could effectively distinguish ankle joint angle differences of at least 2°.

In addition, matched (left) foot data was consistently underestimated at all target angles except for 0°. At angle 0°, data was almost equally distributed between overestimation and underestimation, with a mean constant error of 0.28° (overestimated).

Participants' matched foot position was most accurate (closest to target angle position) at 0° and 2° and was most precise (least variable) at 2°. Performance only worsened with the increase in angle both ways (towards dorsiflexion and plantarflexion). This suggests that participants proprioceptive acuity was highest when the ankle was in neutral position or slightly dorsiflexed, within the range required to maintain quiet standing. Interestingly, a study by Loram et al. (2007) found a substantial passive internal ankle stiffness (calculated as change in torque per change in rotation) present when the neutral ankle was rotated or perturbed by a small amplitude—as small as 0.03°—towards dorsiflexion. In their study, participants were positioned in a "passive standing" posture—similar to our own set-up—and instructed to remain relaxed in their legs during the experimental procedure. They applied low-frequency passive ankle rotations—using 0.1°/s velocity—to the upright subjects, and the footplates were rotated at a constant speed from -1.5° (toes down) to 5.5° (toes up). The experimenter monitored the left and right ankle torques while visual and audio feedback of the tibialis anterior and soleus obtained through surface electromyography (EMG) helped subjects minimise their muscle activity. It was found that intrinsic stiffness decreased from 101 ± 9% to 19 ± 5% as the size of ankle rotation increased from 0.03° to 7°. This seems to imply that ankle stiffness peaks at neutral angles of the ankle, ones typically associated with quiet standing, and decreases when the ankle is rotated away from this range. Therefore, the question that comes to mind is: Can the enhanced proprioceptive acuity we observed around the neutral ankle position (0° and 2°) be related to this observed significantly higher passive intrinsic stiffness of the ankle when it moves through these same neutral angles? At present, no studies could be found on the direct effect of intrinsic ankle stiffness on proprioceptive acuity. However, another explanation for the enhanced proprioceptive performance we observed at 0° and 2° is possibly an optimum level of force (passive torque in this case) going through the muscle spindles at these angles. Several studies have shown that increased spindle activity due to voluntary contraction was associated with enhanced proprioceptive acuity (Vallbo, 1970; Kakuda and Nagaoka, 1998). However, voluntary muscle contraction, although associated with an increase in force within the muscle, is very different physiologically to passive torque force or intrinsic ankle stiffness. Moreover, we observed an effect of age on AE scores specifically at these same two angles 0° and 2° indicating that older participants tended to perform worse at the ankle joint angles typically associated with quiet standing balance. This may explain the increase in risk of falls seen in older individuals and discussed previously in our literature review. Therefore, additional research is needed to understand the relevance of these ankle joint angles and the mechanism that enhances proprioception when the ankle is positioned into neutral.

3.5.2 Ageing effect

Our data indicated that age correlated directly with mean AE and post hoc analyses indicated this effect to be specifically significant at angles 0° and 2°. As discussed in our literature review, age seems to influence muscle spindle efficiency, due to changes in the muscle framework, such as altered fibre length and changes in tendon compliance (Proske and Gandevia, 2012). Moreover, there is evidence that muscle spindles themselves undergo significant structural and functional changes with advanced age: On the structural level, muscle spindles in aged humans possess fewer intrafusal fibres, an increased capsular thickness, and some spindles which show signs of denervation (Kröger and Watkins, 2021). In addition, electrophysical studies showed that mature rat muscle spindles exhibited a lower dynamic response of primary endings compared with those of young animals (Kim et al., 2007). Taken together, this seems to indicate that the proprioceptive system undergoes significant changes associated with advancing age, and that a gradual decline in proprioceptive function in elderly individuals is inevitable. In relation to our own study, this also may implie that in quiet standing—with the absence of other proprioceptive senses (vision, vestibular influence, conscious muscle tension)—age may be a good predictor of ankle joint proprioceptive acuity. This would be in agreement with most of the literature (Gilsing et al., 1995; Hurler et al., 1998; Verschueren et al., 2002; Adamo et al., 2007; Kalisch et al., 2012; Ribeiro and Oliveira, 2011; Yang et al., 2019). On the other hand, the study by Djajadikarta et al. (2020) found that age had little effect on proprioceptive acuity, as measured via three methods of proprioceptive assessment— TTDPM, proprioceptive reaction time, and JPR—when thixotropic muscle history was controlled for. However, the method they used for controlling movement history of the ankle was similar to our own method: Before each target movement, the footplate was repeatedly moved up and down as the foot rested on the footplate. This "waggle"—the term that was used in their study—seems identical in principle to our own two-second "wobble". However, our findings suggest that joint position matching is in fact affected by age when average AE data was analysed. When we analysed data based on age group—separating participants into young, middle-aged, and old age groups— we found that the old age group tended to score the worst in accuracy and precision in joint matching among the three groups. Although —statistically—there was no significant difference between age groups, this may be due to the smaller sample size when data was analysed in this way. According to a power analysis using G*Power (Faul et al., 2007) a sample size of 30 participants would have been needed for a significant result using our analysis parameters. Nevertheless, a large study by Yang et al. (2019) on the influence of age on proprioceptive acuity found that older age had a negative effect on ankle proprioception. Using the AMEDA (Active Movement Extent Discrimination Apparatus) method in standing as their proprioceptive assessment method, they found significant differences between age groups (n = 118, $p = .002$), with young adults (18-25 years (yrs.) showing significantly better proprioceptive acuity than children (6-8 yrs.), adolescents (13-15 yrs.), old adults (60-74 yrs.) and very old adults (75-90 yrs.). More importantly, the performance of very old adults was significantly lower than the other 5 groups (all $p < 0.01$). Although a different method was used to assess proprioception in that study, our results echo theirs in that young adults showed the best proprioceptive acuity, and oldest participants showed the worst. As for the findings of Djajadikarta et al. (2020), although we cannot explain why age had a negligible effect on proprioception, their results suggest muscle *condition* significantly influenced proprioceptive performance, and that by controlling the history of muscle movement, muscle spindles performed similarly in young and older aged participants during their study.

3.5.3 Strength effect

Leg strength correlated inversely with age which would be expected given the effects of sarcopenia associated with age (Vandervoort and McComas, 1986; Proske and Gandevia, 2012). In terms of matching ability, leg strength was shown to exert a significant negative effect on AE scores at the end range angles towards dorsiflexion: **4°** and **6°**. It is well established that when muscles are stretched, muscle spindle firing rate proportionally increases in frequency (Kröger and Watkins, 2021). Does this indicate that stronger muscles equate to better matching ability when said muscle is in a stretched position? No effect of strength was observed at any of the other target angles including the end ranges chosen towards plantarflexion. However, typical dorsiflexion range of motion (M = 30 ± 6.7) constitutes approximately half of the range of motion of plantarflexion (M = 59 ± 8.6). Thus, the end range angles chosen towards plantarflexion (-4° and -6°, approximately a 1:10 ratio of the full range) likely do not exert the same stretch effect on the tibialis anterior muscle as that exerted on the gastrocnemius during 4° and 6° dorsiflexion (approximately a 1:5 ratio of the full range). In addition, when referring back to the passive torque values that were recorded during testing (Appendix 4), we observed that torque increased through the legs with an increase in angle towards dorsiflexion. It may be that stronger lower limb muscles are able to perform more efficiently—in terms of proprioceptive performance—than weaker muscles when a force is exerted through the legs.

Furthermore, several studies indicate muscle strength may have a positive effect on proprioceptive ability. Butler et al. (2008) found that muscle weakness among participants was associated with poor proprioceptive postural control with eyes closed and suggested a relationship between contractile and sensory properties within skeletal muscles. In addition, Hurler et al. (1998) found moderate but significant correlations between quadriceps weakness and decreased joint position acuity, whereas Lordet al. (1993) indicated that muscle strength was among the best sensory-motor predictors of body sway on an unstable surface. Additionally, Ribeiro and Oliveira (2010) reported that although age was shown to decrease position sense ability, regular exercise was shown to compensate for this age-related decline, with no significant difference in AE scores between non-exercised young participants and exercised old participants.

As regular exercise is associated with improved muscle strength, faster reaction times, and improved stability (Lord et al., 1993) it may be safe to assume that stronger more efficient muscles would facilitate the neural pathways involved in proprioception, resulting in improved proprioceptive acuity of the associated joint.

In our study, data was further analysed based on strength level between age groups. Higher strength groups performed better for the middle-aged and old age groups, showing lower mean AE. Nevertheless, the difference in mean AE between higher and lower strength participants was only significant for the middle-aged group.

However, an interesting trend was apparent between all age groups: Lower strength participants tended to have lower AE scores at neutral/mid-ranges used in the experiment while higher strength participants showed lower AE scores toward the end ranges both ways (Figure 4.10). It may be that muscle strength is associated with improved proprioception only when contractile tissue is somewhat stretched, stimulating sensory receptors. However, the higher strength group tended to perform worse at neutral ranges. More studies need to be conducted to clarify this association.

3.5.4 Limitations

An obvious limitation of our study was that the experimental setup (standing supported by a semi-vertical backboard) is comparatively different to actual quiet standing. As participants were not burdened with maintaining their own balance via continuous muscle activity of the anterior tibialis and calf muscles typically present during quite stance (Loram et al., 2007), it is possible that their proprioceptive performance was altered (either enhanced or diminished). However, by removing this motor output it was likely we were better able to assess the ability of peripheral sensors (muscle spindles) to detect changes in foot position.

Another notable limitation was our positioning during the MVC assessment of the calf muscles (gastrocnemius and soleus). While theoretically the optimal knee joint angle for maximum muscle recruitment of both the gastrocnemius and soleus is 180°, or slightly bent at approximately 160° as demonstrated by Trappe et al. (2001), in our study participants were seated with their knees flexed approximately 85° from the horizontal. In this position the gastrocnemius would be slightly slack, as it is a bi articular muscle, and thus unable to generate maximum muscle force. However, we believe in flexing the knee we were better able to isolate the plantarflexion action from any other muscle force that may be generated in a more extended knee position. Additionally, the MVC score was used as a reference to compare muscle strength between participants in this study and therefore seems adequate for our research needs.

3.6 Conclusion

The current study aimed to assess whether muscle strength was directly related to proprioceptive ability, and whether age-related decline in proprioceptive ability was counterbalanced by an individuals' muscle strength.

Our results suggest that age does indeed affect proprioceptive ability, with older participants performing worse at the matching task than younger participants. In addition, while muscle strength does seem to influence proprioceptive ability, where higher strength seemed to be associated with lower AE scores for both the middle-aged and old age groups, this was only significant for the former age group. Regarding specific target angles, strength had a significant negative effect on AE only for the end range angles towards flexion (4° and 6°), and higher strength individuals showed better proprioceptive ability than lower strength individuals at these ranges.

To conclude, we hypothesised that age would influence proprioceptive ability at the ankle, and that this effect would be related to changes in muscle strength. We found that while age does influence proprioception, it is less clear how strength effects this proprioception precisely, as our results were not so clear cut. Further studies are needed to clarify the effect and conditions in which muscle strength effect proprioception.

Chapter 4: Active and Passive Detection of Threshold

Abstract

For our second experiment, we aimed to determine whether isometric contraction influences proprioceptive acuity at the ankle joint, as measured by threshold to detection of passive motion (TTDPM). A total of 30 healthy young subjects with a mean age of 24.4 ± 5.9 years participated in the study. Proprioception was evaluated using the first of our developed TTDPM methods, in which detection threshold is measured as angular amplitude (°) at the moment of movement detection. Participants' detection threshold was measured during two conditions: an active condition, in which they were required to maintain an isometric contraction at their calf-muscle throughout testing, and a passive condition, in which they were asked to remain relaxed and passive in their legs. Our main outcome measures were detection threshold amplitude (°) and torque variability (N.m). Our results indicated no significant difference in proprioceptive acuity between the active and passive conditions ($p=0.2$). However, we found a significant positive correlation between torque variability and detection threshold scores ($r_{(29)}= 0.47$, $p< 0.001$). When we separated our data based on testing condition, we found that torque variability significantly predicted detection threshold scores for both the active ($r_{(29)}= 0.44$, $p=0.01$) and passive ($r_{(29)}= 0.47$, $p= 0.01$) conditions. This suggests that the steadier a participant is able to maintain a constant force, the better they are able to detect footplate movement.

4.1 Introduction

Understanding the mechanisms through which the body maintains standing balance in healthy individuals is paramount in developing effective rehabilitation programs for clinical populations with balance problems. Studying the effect of muscle strength on one's proprioceptive acuity, as we did in the previous chapter, is but one aspect of a multidimensional process. Another aspect we planned to research is the effect of voluntary muscle contraction on proprioceptive acuity. Instead of measuring maximum voluntary contraction (MVC) values to use as a separate outcome measure in our analyses, as we did in the previous chapter, for the present chapter we had participants voluntarily contract their muscle *while* performing a proprioceptive task to analyse whether proprioceptive acuity was affected by muscle condition. Three types of voluntary muscle contraction are known at present, and all three rely on varying levels of tension and length. Whereas muscle contraction always involves an *increase* in tension of the muscle, this is not always synonymous with shortening of muscle length (otherwise known as *concentric contraction*), as tension within the muscle may be produced while muscle length stays intact (*isometric contraction*) and while it is lengthened (*eccentric contraction*) (Gash et al., 2022).

4.1.1 Literature review

With regard to standing balance, forward collapse of the body is prevented by ankle torque, which is produced by activity of the triceps surae muscles. The activated triceps surae muscles generate an intrinsic mechanical stiffness across the ankle joint, and such stiffness provides an instant torque response to any change in ankle angle (Loram and Lakie, 2002). By delivering alternating impulses via the gastrocnemius and soleus muscles, our nervous system is able to effectively regulate centre of mass (CoM) movement to control quiet standing balance, while the bias adjustments themselves result in what is observed as postural sway during standing (Loram et al., 2005). Interestingly, it has been reported repeatedly in the literature that a level of muscle contraction improved positional acuity. Several groups found that subjects' joint position sense was somewhat poor when their limb was passively positioned (Goodwin et al., 1972; Gregory et al., 1988) compared to when they were asked to actively position their limb (Laufer et al., 2001; Paillard & Brouchon, 1968). A possible explanation for the improved performance would be that increased spindle activity due to voluntary contraction (Kakuda & Nagaoka, 1998; Vallbo, 1970) provided more precise proprioceptive information. Consistent with the previous explanation is a a study by Suprak et al. (2010), in which 25 participants were asked to perform a joint position reproduction (JPR) task during five different conditions of external resistance: an unloaded condition and four varying external load conditions imposed by added wrist weights. Results indicated that JPR errors decreased linearly as the external load increased up to 40% above unloaded shoulder torque ($p=0.019$). This finding suggests that JPR accuracy improves as muscle activation levels—and consequently muscle spindle activity—increase. In addition, a study by Winter et al. (2005) found that participants were significantly more accurate during a proprioceptive matching task at the elbow when they voluntarily contracted their muscles (unsupported condition) to reposition their forearm compared to when the forearm was repositioned passively (supported condition) ($p<0.01$). However, when an external load of 2 kilograms was added to the forearm (weighted condition), position matching did not improve. Although the difference in joint position error was significant between the unsupported and weighted conditions ($p<0.001$), error *values* were not improved, only made in different directions. Errors made during the weighted condition tended to be in the direction of extension relative to the errors made in the unsupported condition. If increased spindle activity was indeed equated with improved proprioception, then joint matching accuracy should have further improved as more spindles became coactivated. However this was not the case here. Moreover, when muscle flexion and extension conditioning was performed prior to testing, it was observed that the difference in position error *direction or bias*—between flexion conditioning and extension conditioning—was significant for all three testing conditions (supported, unsupported and weighted). However, the average size of this difference fell from 4.3 ± 1.1° (supported condition) to 3.3 ± 0.6°(unsupported condition) to 1.9 ± 0.4° (weighted condition). This seems to imply that the position signal was coming from passive muscle spindles, as conditioning-dependent changes were most prominent for the passive supported condition, and least prominent in the weighted condition. These findings also suggest that when muscle spindles are activated through the fusimotor system, they no longer contribute to position sense. However, the improved JPR accuracy during the unsupported versus the supported condition indicates that an effort signal generated during motor command may provide some type of additional feedback on joint position.

On that note, several more recent studies have shown that voluntary muscle contraction worsened proprioceptive acuity. For example, Niessen et al. (2009)

investigated the differences between active and passive JPR at the shoulder, and found significantly larger errors during active JPR compared with passive JPR (*$p<0.01$*). Subjects also were significantly less consistent during active JPR (*$p<0.01$*). They concluded that movement mode (active or passive) significantly influences proprioceptive errors and proprioceptive consistency. Also, a cross-sectional study by Kwon et al. (2013) investigated the effect of repetitive passive and repetitive active movement on active and passive JPR during forearm supination. Active and passive JPR was measured both before and after receiving the repetitive passive and active movements. Results showed that in the repetitive passive movement test, there was a statistically significant *decrease* in the pre- versus post-repositioning error scores in both the active (*$p=0.041$*) and passive (*$p=0.041$*) JPR assessments. Whereas in the repetitive active movement test, there was a statistically significant *increase* in pre- versus post-repositioning error scores in the active (*$p=0.008$*) and passive (*$p=0.039$*) JPR assessments. In addition, when comparing JPR sense *between* repetitive movement conditions, it was found that both active (*$p=0.001$*) and passive (*$p=0.002$*) JPR error decreased significantly during the repetitive **passive** movement condition when compared to the repetitive active movement condition. These results indicate that repetitive passive movements seem to result in greater proprioceptive acuity, and could be a better option when selecting rehabilitation exercises in clinic for populations with proprioceptive deficits.

Similar to these results were those from a study by Peters et al. (2017), in which it was found that active muscle contraction reduced muscle spindle sensitivity. In an attempt to understand precisely just how muscle spindles contribute to the control of balance, they used microneurography to record axon potentials directly from single human muscle spindle afferents in the triceps surae muscles during servo-controlled movement of the ankle joint. Subjects were made to lay prone and completely passive with their feet placed in custom-built ankle platforms designed specifically for microneurography experiments on the tibial nerve. To simulate movements of the ankle while standing once the electrodes were in place, they delivered random 90 second dorsiflexion/plantarflexion oscillations of the ankle joint, with a peak-to-peak amplitude of 0.7° and frequency content below 0.5 Hz. After the subjects were tested passively, they were instructed to hold a low-level near-isometric plantar flexion contraction, which equated to ~2-7% of maximal voluntary contraction. It was demonstrated that afferent activity of these muscle spindles closely reflected ankle movements at amplitudes and frequencies characteristic of human standing. Additionally, it was found that although muscle spindles of both the gastrocnemius and soleus were sensitive enough to code for the ankle movements associated with quiet standing, the soleus had the most sensitive muscle spindle afferents at all stimulus frequencies. As for the voluntary contraction of the triceps surae, it was observed that it *reduced* spindle sensitivity, but this was only significant near the mean power frequency of the stimulus (~0.3Hz) ($p < 0.05$).

All the previously mentioned studies tested proprioceptive acuity as JPR. And so naturally a question that comes to mind is: What effect would active contraction of the calf muscle have on proprioceptive acuity measured via threshold to detection of passive motion (TTDPM)? Would it heighten or dampen our sensitivity to detect ankle motion?

This is different to the concept of thixotropy, which refers to the passive property of muscles in which the discharge of muscle spindle endings is affected by previous stretch or by previous fusimotor activation during voluntary contraction of the receptor-bearing muscle. It is well established that thixotropy has a significant effect on proprioception, with the change in discharge rate after a stretch or a voluntary contraction affecting a person's perception of joint position (Macefield & Knellwolf, 2018; Proske & Gandevia, 2012).

Our query about the effect of voluntary contraction on proprioception is also distinct to the concept of muscle fatigue following exercise, which has been shown to disturb proprioceptive acuity by influencing the afferent spindle receptors in the muscle (Lee et al., 2003). Although all three types of exercise (concentric, eccentric, isometric) can cause muscle fatigue if carried out sufficiently, it has been shown that concentric exercise disturbed position sense to a greater degree than eccentric or isometric exercise (Proske & Gandevia, 2012). Also, it has been claimed that TTDPM at the shoulder is significantly increased by fatigue from exercise (Carpenter et al., 1998).

Moreover, what we mean by active contraction in our question above is a sustained, isometric contraction of the calf muscle during a proprioceptive task, in this case TTDPM at the ankle joint, during standing. As of yet, there is no consensus on how exactly muscle spindles respond to voluntary contraction, as this response seems to be relative to not only the type of contraction, but also to the muscle group involved, the task (or lack thereof) at hand, the positioning of the subject during the contraction, and for how long the voluntary contraction lasts (Macefield & Knellwolf, 2018). In general, it has been observed that human muscle spindles in agonist muscles lose their sensitivity to length changes during concentric contractions (Jahnke & Struppler, 1989; Peters et al., 2017). However, this seems to be compensated for by the length and velocity information encoded by muscle spindles in the antagonist muscle as it lengthens (Dimitriou, 2014). As for isometric contraction, it has been shown that when the agonist and antagonist co-contract, spindle firing rate is higher than it is in the previous two types of muscle contraction. It is thought that this increased discharge rate of muscle endings may be due to either the smaller length changes characteristic of isometric contraction, and to which muscle spindles are more sensitive, or to a disproportionately high fusimotor drive during co-contraction (Nielsen et al., 1994). Having said that, it would seem that muscle spindles are at their most sensitive when the muscle is in a relaxed state (Winter et al., 2005; Niessen et al., 2009; Peters et al., 2017).

How would this concept apply when a subject is in a quiet standing position though?

Referring back to our experimental setup (which was used for all experiments included in our research) the subject is in a bilateral weight bearing position; however the use of a backboard meant that muscle activity in the lower limbs was at a minimum, but enough to be able to maintain a supported standing position. Would adding an isometric contraction to the calf muscle (triceps surae) enhance proprioceptive acuity of the ankle in standing in this case? Or would remaining passive result in better ankle proprioception?

To the knowledge of the researcher, no study to date has researched the effect of a sustained isometric contraction on one's ankle proprioceptive acuity while performing a TTDPM task.

4.1.2 Aim of our study

Our study aimed to test whether isometric contraction influenced proprioceptive acuity at the ankle joint. We determined the effect of holding an isometric contraction while performing a proprioceptive task during standing and established whether any one of the muscle conditions (active or passive) is superior to the other in terms of proprioceptive acuity.

To do this, proprioception was evaluated through TTDPM, using a method we developed specifically for our setup (see Chapter 2). It involved numerous testing sessions using the staircase method, a psychophysical process which is used to locate proprioceptive threshold for a specific parameter. In our case, we searched for the detection threshold, or in other words, the amplitude at which a subject achieved approximately 80% performance level or perceived the stimulus correctly 80% of the time (see Chapter 2-section 2.4.3). This method was tested at the ankle joint during two different muscle conditions:

-Active: Subject maintained an isometric contraction for the duration of testing

-Passive: Subject remained relaxed throughout the duration of testing.

4.1.3 Hypotheses

We hypothesised that sustained isometric contraction will significantly affect proprioceptive acuity as measured by TTDPM.

Null Hypotheses (H_0):

-Contraction of the calf muscle will have no significant effect on proprioceptive acuity at the ankle joint.

Alternative Hypotheses (H_a):

-Contraction of the calf muscle will have a significant effect on proprioceptive acuity at the ankle joint.

The current study will determine whether there is a relationship between active muscle contraction and proprioceptive acuity at the ankle during standing, and if so, whether proprioceptive performance is improved or worsened by this contraction.

4.2 Methods

Before starting our experimental trials, we conducted a separate preliminary study to investigate the average ranges of ankle proprioceptive detection threshold in the population using different velocities. The velocities chosen were 0.1, 0.25 and 0.5 °/s. Throughout this chapter, sections will be divided to include both studies.

4.2.1 Participants

For the preliminary study, ten healthy young adults (6 female) with a mean age of 22.9 ± 4.95 years participated. As for the experimental study, thirty healthy young adults (19 male) with a mean age of 24.4 ± 5.9 years participated. Participants were recruited via email between February 2020 and June 2021. Inclusion criteria for both experiments were as follows: Healthy and aged between 18-65 years; exclusion criteria included any ongoing physical or neurological injury or illness which might affect normal control of standing balance and/or any injury to the lower limbs. The experiments took place in the Human Balance Laboratory in the School of Sports, Exercise and Rehabilitation in the University of Birmingham, and each took approximately 1.5 hours to complete per participant. The experiments were approved by the School of Sport, Exercise and Rehabilitation Ethical Review Committee at the University of Birmingham (ERN_14-0740A) and performed in accordance with the Declaration of Helsinki (World Medical Association, 2013). Participants were given information sheets on the experiments and provided written, informed consent before starting the trials.

4.2.2 Experimental setup

For both studies participants stood barefoot with their back leaning against a board angled 21.8° from vertical, with a Velcro strap fastened around the lower thighs to minimise any movement of the knees. A foam cushion was placed behind both knees for comfort. See Figure 4.1. This setup was chosen to ensure the person was bilaterally weight bearing without the need to balance themselves as they would when standing normally. It also allowed them to maintain a standing position using minimal muscle activity without collapsing. Each foot was placed on a separate motorised footplate, which could be rotated individually using two linear motors attached by levers (Model XTA3810S; Copey Motion Systems LLC, Basildon, UK). The axis of rotation of each footplate was made to be approximately collinear with the ankle joint. In addition, each foot plate was affixed with sensors for position (Model CP-2UT; Midori Precisions Co., Tokyo, Japan) and torque (Model 31, Sensotec Inc., Columbus, OH, USA).

Figure 4.1: Participant positioning while performing the detection of threshold task. Standing with their arms by their sides or, alternatively, crossed in front of their chest, participants were asked to keep their eyes closed and their body completely relaxed during testing, except for their lower extremities during the active testing condition.

4.2.3 Procedure

Preliminary study: To assess ankle proprioceptive detection threshold for each of the three chosen velocities (0.1, 0.25 or 0.5 degree per second (°/s)), participants completed one trial per velocity, with the length of the trial dependant on participants' individual performance. Each trial consisted of an un-fixed number of movement sequences, and the amplitude of movement was adjusted after each individual movement sequence in a staircase-like manner. This will be explained in further detail in the *scoring* section. The footplate movements were generated through MATLAB (MathWorks Inc., Natick, MA, USA) to move the **right** footplate either up (dorsiflexion) or down (plantarflexion) from the start position of 0^{o} (perpendicular to angled backboard). The **left** footplate remained at the reference 0^{o} start position during the entire experiment.

The movement sequences referred to above were at most three consecutive movements per sequence, consisting of footplate movements either upward or downward in random order (eg. Up up down, down up down, down down down) and were separated by a two-second period of random dorsiflexion and plantarflexion motion (the same wobble we used during our joint matching experiment in Chapter 3) before settling at the start position (0°) before the next movement sequence started. The purpose of the motion was to wash out any history-dependent effects between movements (Proske et al., 1993).

The position trace of a typical experimental movement sequence is presented in Figure 4.2. The 3 consecutive movements per sequence can be seen clearly in the figure (up, up, down). Movement direction was randomised for each trial. This trace represents the movement of the right foot pedal for a single sequence within the experiment.

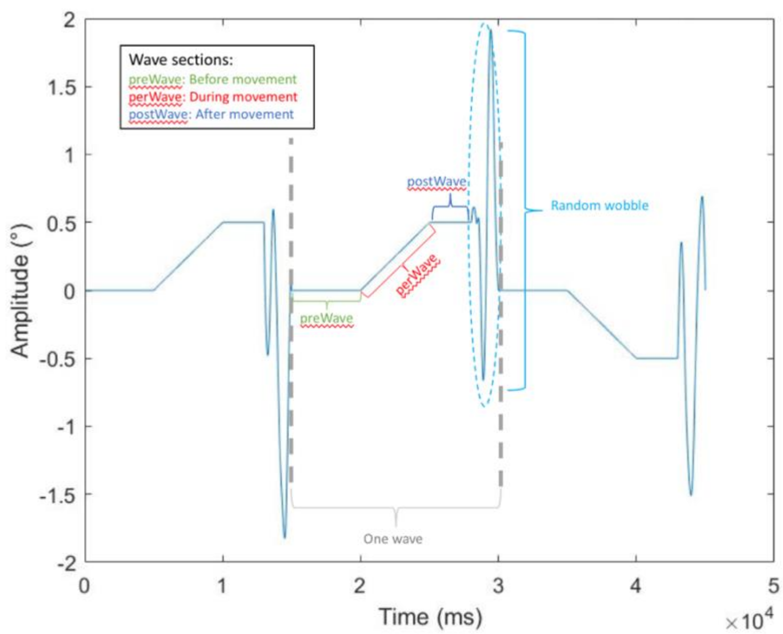


Figure 4.2: A position trace for the right foot pedal during a typical experimental trial. This trial consisted of three "waves" or movements of the foot pedal: The middle waves has been annotated to clarify the four sections of a typical wave (preWave, perWave, postWave and random wobble). This same wording was used to label these sections of recorded data during the experiment.

During the experiment, participants (with their eyes closed) were asked to report whether each movement was "up" or "down". A series of beeps were generated in parallel with the movement sequence, to alert the participant of the following:

1. Three short beeps that finish 0.5s before the sequence starts. This signalled to the participant that the trial was about to begin.
2. A long beep time-locked to the duration of motion of the footplate. This signalled to the participant that the footplate was moving.
3. A short single beep at the end of the sequence. This signalled to the participant that the sequence had ended. At this point they were required to report whether they felt the footplate move up (dorsiflexion) or down (plantarflexion).

Experimental study: For the experimental study, the procedure was repeated precisely the same as for the preliminary study; however only one velocity—0.1 (°/s)—was chosen for testing.

4.2.4 Scoring

During the movement sequence for both experiments, the participants were asked whether they felt their right foot had moved "up" or "down" when they heard the short beep indicating the end of the movement. Depending on their answer, the sequence could end after one movement, two movements or it could continue to complete three movements.

A staircase-like psychophysical method was adapted to determine ankle proprioceptive detection threshold following a three-down one-up algorithm combining transformed and weighted staircase procedures which we developed (described in Chapter 2). This was done to produce a more efficient staircase method, as traditional procedures have been shown to be time-consuming with insufficient precision (García-Pérez 1997).

With the three-down one-up rule applied, if the participant were to answer correctly three times in a row (thus the three-movement sequence) the amplitude would be decreased (step down -Δ) and if they were to answer incorrectly once, the amplitude would be increased (step up +Δ), with the sequence ending at the incorrect answer (either after the first, second or third movement in the three-movement sequence). In line with the recommendations of García-Pérez (1998), we used a larger step size up than down. This was achieved by halving the stimulus (amplitude of rotation (°)) at *every other* step as follows (Figure 4.3):

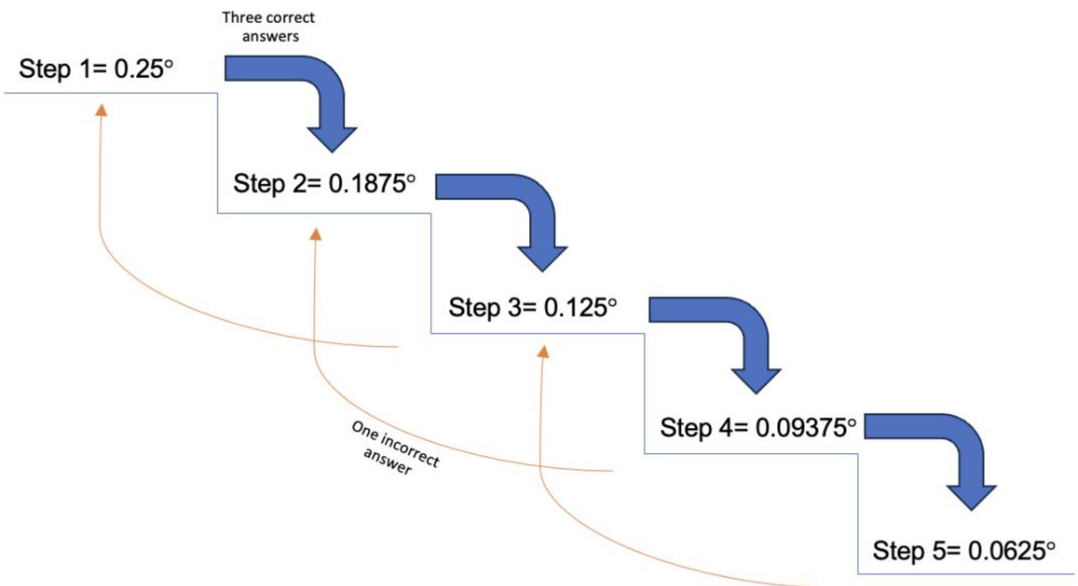


Figure 4.3: Stimulus "steps" used in our detection threshold experiment. Assuming we started the experiment at an amplitude of 0.25° (step 1) the subject would need to correctly perceive the direction of movement (up or down) three times at this amplitude in order to move one step down (to step 2: 0.1875°). They would again need to answer correctly three times in order to move to the next step (step 3: 0.125°) and so on. If the subject was unable to give three correct answers, the would be moved up to the step before last, so amplitude would be doubled. The amplitudes chosen for each step were such that for every *other* step the amplitude would be halved (going down) or doubled (going up).

For our staircase, a total of 8 reversals in direction were required to end testing. However, attaining 8 reversals meant that the number of trials was *not* fixed, and differed depending on participants' performance. As such, trial time was inconsistent between participants, ranging anywhere from 45 minutes to 1.5 hours.

After the participant reached 8 reversals, the average of these reversal points (amplitude°) was measured, and this value was considered the subject's detection threshold.

<u>4.2.5 Active and Passive conditions</u>

During the experimental study, using the above procedure, each participant completed two phases, an active and passive phase. The order of these two phases was alternated for each consecutive participant (active-passive or passive-active).

However, before determining detection thresholds, the participant, while standing in place, was asked to perform a series of tasks to obtain specific "reference" torque values to be used during the trials. These values were related to the right footplate torque, which were sampled at 1000Hz using Simulink Real-time Desktop (MathWorks Inc., Natick, MA, USA) and consisted of the following:

<u>Baseline:</u> Right foot is lifted completely off the right plate to determine the baseline torque due to the weight of the footplate alone.

<u>Passive:</u> Participant stands entirely relaxed with both feet in place. They are told to let their legs be entirely limp, such that no muscle activity occurs (The knee strap prevents the leg from buckling).

<u>Heel-off:</u> Participant is asked to lift *left foot* completely off the left foot plate while raising only the ***heel*** *of the right foo*t approximately 1-2 cm off the right foot plate. This was done to obtain a reference voluntary contraction (RVC) of

the gastrocnemius muscle to standardise the force used in the active trial task between participants.

Using the heel-off value, it was possible to calculate each participant's 5% and 15% RVC, creating a range for the force the participant needed to push with in the active condition. The task was to maintain the torque between the two values (5% and 15% RVC) with a rough target of 10% RVC of force throughout the movement sequence. It did not demand high precision, so therefore was not an attentionally demanding task. A practice trial was performed after obtaining these values where the participant was asked to push with a force within the 5% to 15% range–which was presented visually on a screen in front of them via a Simulink signal trace–and attempt to maintain this force with their eyes closed. When the participant was happy with their performance the experimental trial would begin. To measure participants torque and torque variability (which refers to the standard deviation of their torque) throughout the experiment, the Simulink signal trace mentioned above was recorded for each participant during both the active and passive phases of the detection threshold trials.

For the active condition: Participants were asked to close their eyes and press with the same force they practiced previously (between 5% and 15% RVC) before the trial began. They were asked to maintain this pressure during the entire movement sequence for testing and were told when they could relax at the end of the sequence. The researcher was able to observe the force with which the participant was pressing in real time via a torque trace only visible to the researcher. In this way it was possible to insure that participants were indeed pressing through their leg to the required 5-15% RVC range throughout the active testing condition. If participants were to deviate from this range, they were notified by the researcher to adjust their pressure accordingly. In summary, the procedure for the active condition was the same as it was for the preliminary study except for the addition of the self-produced force through the participants right leg while testing.

For the passive condition: Participants were asked to close their eyes and remain as relaxed as possible during the movement sequence, same as the method used for the preliminary study.

4.3 Analyses

In terms of analyses, GraphPad (GraphPad Prism, GraphPad Software, Boston, Massachusetts USA) and SPSS software (IBM SPSS Statistics, IBM Corp, Armonk, NY, USA) (Prism) were used to analyse the data. Descriptive statistics were performed on all data to get an overall picture of our results. We then conducted a Shapiro-Wilk test of normality which indicated that our data was non-normal for both the preliminary and experimental studies ($p<0.05$), and so we used non-parametric tests such as the Friedman test (the non-parametric version of a repeated measures ANOVA) to measure the effect of velocity on threshold scores, a Wilcoxon signed-rank test (the non-parametric version of a dependent t-test) to analyse the difference in detection threshold, as well as joint torque and joint torque variability, between active and passive conditions, and a Spearman correlation (the non-parametric version of a Pearson correlation) to analyse the relationship between threshold and torque data.

4.4 Results

4.4.1 Preliminary study

Data from the preliminary detection threshold experiment will be presented under the following subcategories:

1. Demographic data
2. Descriptive and inferential statistics

1. Demographic data:

A total of 10 healthy subjects (6 male, 4 female) with a mean age of 22.9 ± 4.95 years were included in this study. Participants average height was 170.3 ± 13.82 cm, average weight was 67.86 ± 18.58 kg and average BMI was 23 ± 3.1 kg/m^2.

2. Descriptive and inferential statistics:

Three different velocities were tested during the preliminary study: (from the slowest) 0.1, 0.25 and 0.5 °/sec. Descriptive results of our data indicated that detection threshold decreased with the increase in velocity, with an average detection threshold of 0.76° ± 0.71°, 0.35° ± 0.35° and 0.18° ± 0.1° for the velocities 0.1, 0.25 and 0.5°/s respectively (Figure 4.4). To check whether the difference between detection threshold means was significant, we conducted a Friedman test, which indicated that detection threshold scores were indeed significantly different between velocities ($\chi^2_{(2)} = 12.6$, $p = 0.022$). This suggests that velocity had a significant effect on detection threshold, in what appears to be an inverse relationship between velocity and detection threshold scores.

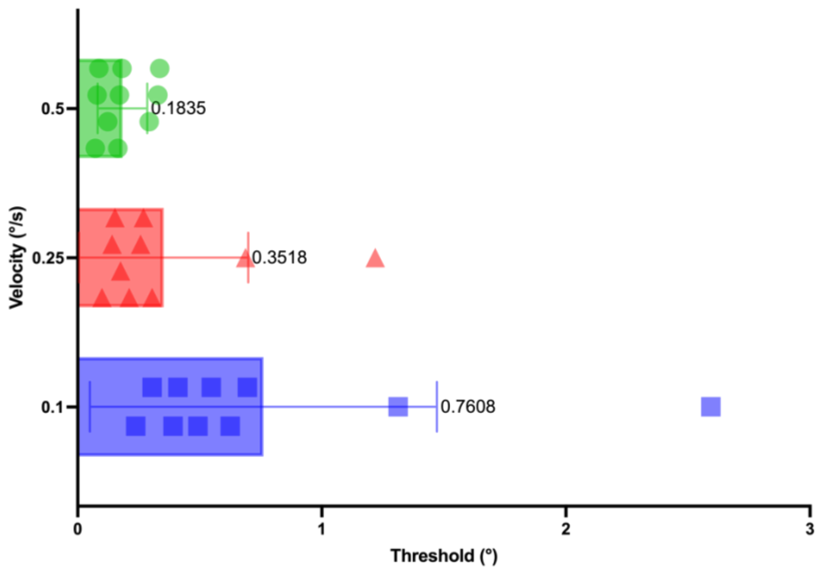


Figure 4.4: Individual detection threshold scores during our preliminary study. There seems to be a clear inverse relationship between velocity and detection threshold scores, with the mean detection threshold (value beside bar) decreasing as velocity increases.

4.4.2 Experimental study

Data from the passive/active detection threshold experiment will be presented under the following subcategories:

1. Demographic data
2. Descriptive and inferential statistics

a. Detection threshold:

    1. Scoring

    2. Results

b. Torque results.

c. Relationship between detection thresholds and torque.

d. Relevance of starting condition (passive or active) on test results.

e. Effect of testing condition (passive or active) on frequency of incorrect answers during testing.

1. Demographic data:

A total of 30 healthy subjects (19 male, 11 female) with a mean age of 24.4 ± 5.9 years were included in this study. An a priori G*Power analysis indicated that based on a repeated measures comparison, a total sample of 35 subjects would be needed to detect a large effect size ($d = 0.5$) with 80% power ($1-\beta = 0.8$) and alpha (α) set at 0.05. A post-hoc analysis indicated that the current sample size of 30 yields a similar effect size ($d = 0.48$) with power ($1-\beta = 0.73$) ($\alpha = 0.05$), which seems adequate for the current experiment.

2. Descriptive and inferential statistics

a. <u>Detection threshold</u>:

For the experimental study, a velocity of 0.1 °/s was fixed for all trials for the entire experiment. This value was chosen based on variability of test results obtained from the preliminary study (Figure 4.3), as it was shown that for the slowest velocity 0.1°/s, participants tended to exhibit a wider range of detection threshold scores (2.36°) compared to the two faster velocities (1.12° and 0.26° respectively). This larger scatter of threshold values makes it more likely to reveal differences between participants. Although outliers most likely skewed this range slightly, nonetheless, it is clear from Figure 4.4 that velocity 0.1°/s is the most variable among the three velocities.

1. Scoring: Calculating detection threshold

We used a modified three-down one-up staircase to obtain detection threshold for our experiment. This was developed by combining transformed and weighted staircase procedures in a process outlined in detail in chapter two. An example of the staircase method of testing is shown in Figure 4.5. This was for a passive phase of the experiment, and it is apparent in this case it took a total of 42 footplate movements to obtain 8 reversals in direction. The average of these reversal points for this participant was 0.28°.

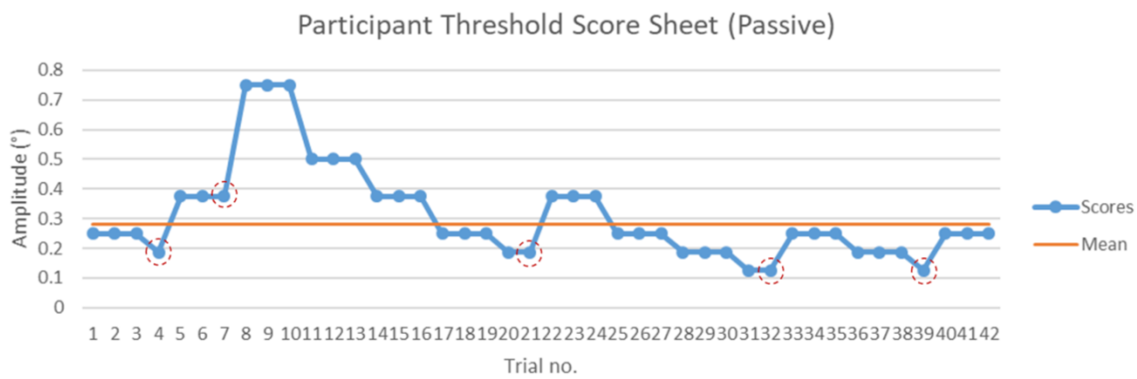


Figure 4.5: Staircase method of scoring to obtain passive detection threshold. It took 42 footplate movements to obtain a total of 8 reversals in direction for this subject. The reversal points here are (0.1875, 0.75, 0.1875, 0.375, 0.125, 0.25, 0.125, 0.25) and the average of these values (or the passive detection threshold) is 0.28°.

An example of the staircase method for an active phase can be seen in Figure 4.6. For this trial it took a total of 50 footplate movements to obtain 8 reversals in direction. The average of these reversal points for this participant was 0.69°.

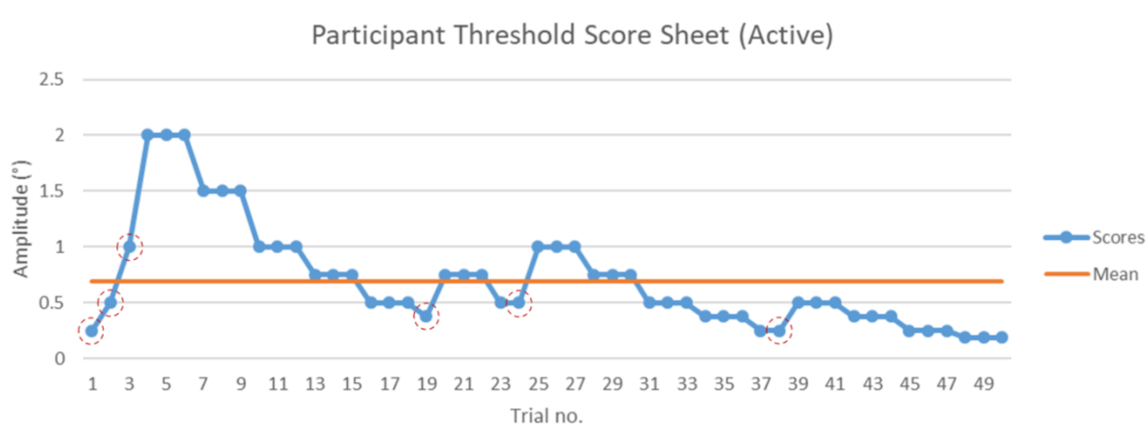


Figure 4.6: Staircase method of scoring to obtain active detection threshold. It took 50 foot plate movements to obtain a total of 8 reversals in direction for this subject. The reversal points here are (2, 0.375, 0.75, 0.5, 1, 0.25, 0.5, 0.1875) and the average of these values (the active detection threshold) is 0.69°.

2. Detection threshold results

Figure 4.7 shows the individual detection threshold scores for each participant, for both passive and active conditions. Upon analysing the data descriptively, average passive detection threshold (0.54° ± 0.26°) appeared to be lower than average active detection threshold (0.68° ± 0.36°). This would suggest that subjects perceived the direction of ankle movement with more accuracy when they were relaxed versus when they were actively pressing against the foot pedal. However, a Wilcoxon Signed-Ranks test found the difference in detection thresholds between conditions to be non-significant (*p=0.2*).

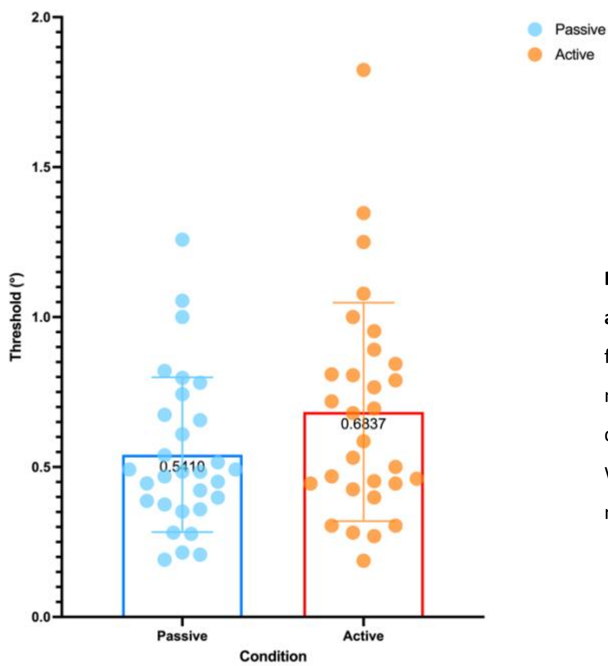


Figure 4.7: Individual detection threshold values for passive and active conditions. Group mean and standard deviation for each of the conditions are presented, showing a greater mean threshold and larger spread of the data during the active condition compared to the passive condition. However, a Wilcoxon Signed Ranks test indicated this difference to be non-significant (p>0.05).

b. <u>Torque results</u>:

In analysing the torque data, we sought to confirm that the signal being analysed reflected what the subject was asked to do during each testing condition (passive or active). For the passive phase the subject was asked to relax and stand like they would normally do with weight distributed evenly between both legs for the whole trial. For the active phase, they were asked to press with a force that was approximately 10% of their individual reference voluntary contraction or “RVC” (refer to methods section for more details on this) also for the entire trial. However, the random wobble that followed each consecutive movement may have distorted torque data so that it did not reflect actual torque. For this reason, time windows from the torque data were selected for analyses to avoid the “wobble” sections. Specifically, two torque sections: “pre-Wave” corresponding to the 3 seconds before the footplate movement occurred, and “per-Wave” corresponding to the time in which the footplate was moving, were analysed for their torque data for each subject. These sections can be seen clearly in Figure 4.2. Studying the torque traces alongside the position traces—Figures 4.8 and 4.9 show an example of this comparison—it became clear that we needed to omit the “per-Wave” section from our analyses as well, as the footplate movements clearly affected torque values during our testing. This meant that for all analyses involving torque data, **we used only the “pre-Wave”**

time window data: so when we refer to torque from here on it may be assumed that this is the torque exerted when participants were either relaxed (passive condition) or pressing against the footplate (active condition) ***before*** any movement of the footplate commenced.

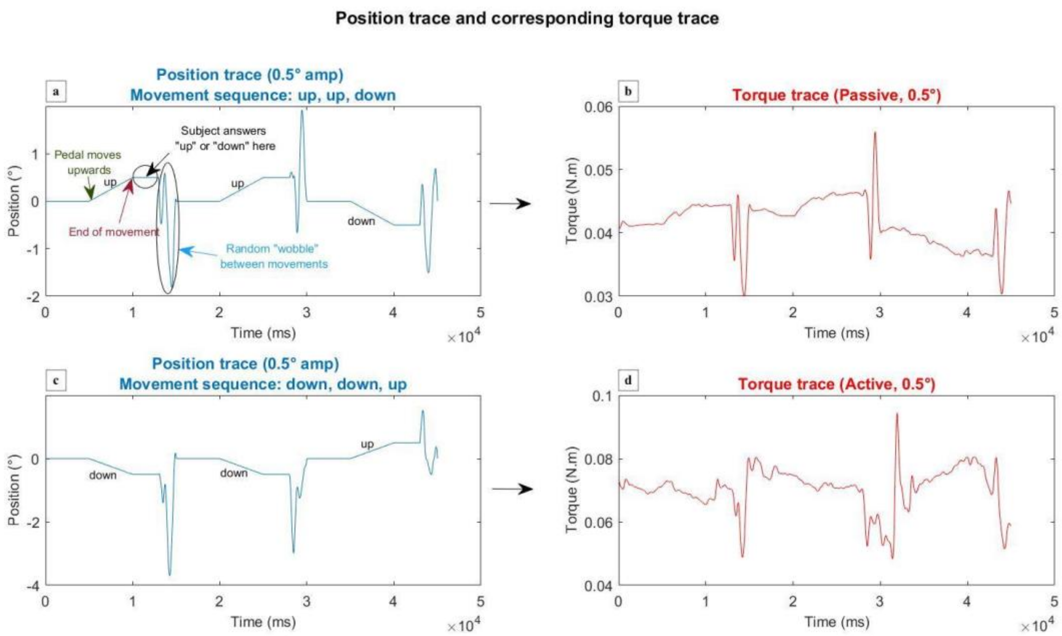


Figure 4.8: Position and torque traces for a passive (a,b) and active (c,d) experimental trial. Figure (4.8a) shows the exact sequence of events during testing: where the foot pedal starts to move, where it stops, when the participant is required to answer whether they felt the movement as "up" or "down" and when the "random wobble" resets the foot pedal in preparation for the next movement. On the right the torque traces for the passive (4.8b) and active (4.8d) conditions can be compared. As would be assumed, torque values are elevated in the active condition, confirming that the participant was indeed pressing and maintaining the force throughout the testing period. This can be seen more clearly in Figure (4.9) below.

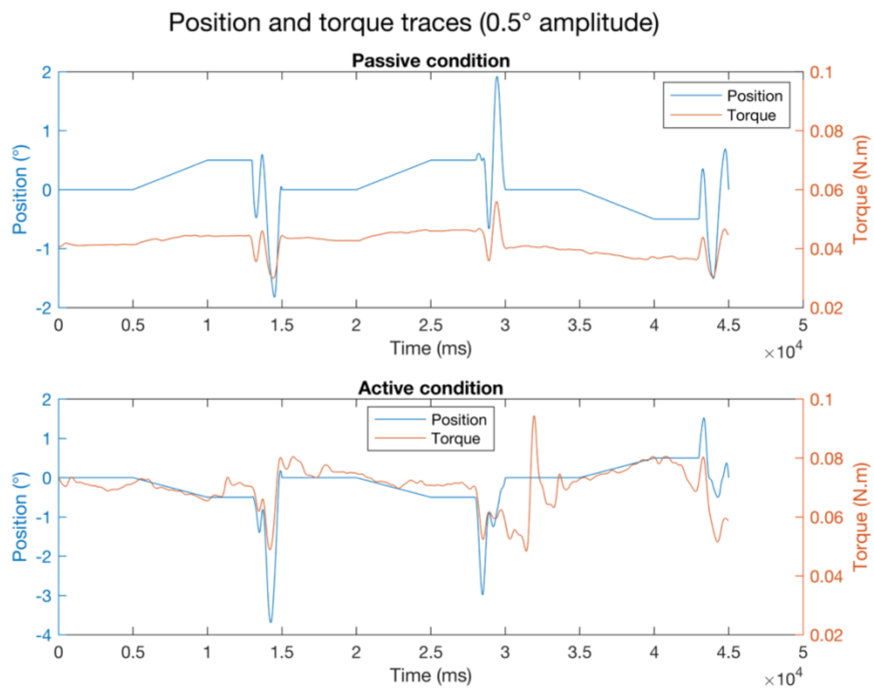


Figure 4.9: Overlapped position and torque traces. These were for the same passive and active experimental trials as Figure (4.8).

In Figure 4.10, a comparison of the individual values for torque mean for each participant for passive and active conditions is shown, along with the group mean and standard deviation for each condition. Torque values appear higher for the active condition, which suggests that participants were effectively holding an isometric contraction during testing. This is also corroborated in Figure 4.11, which shows that all participants recorded higher torques during the active condition than they did during the passive condition. A Wilcoxon Signed-Ranks test confirmed that active torque (M= 0.057 $\pm$ 0.02) was significantly higher than passive torque (M= 0.026 $\pm$ 0.01): (z = -4.78, *p* < *.001*).

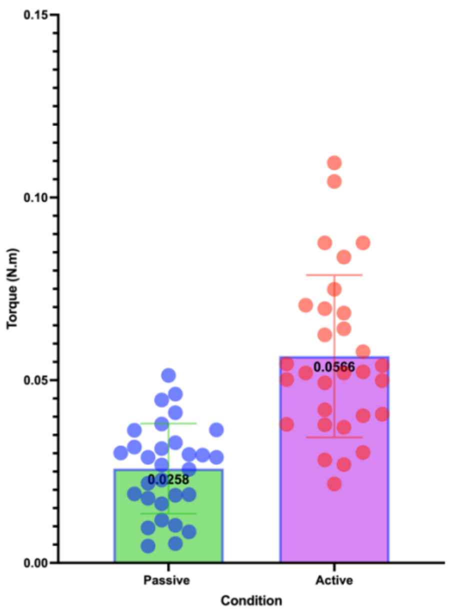
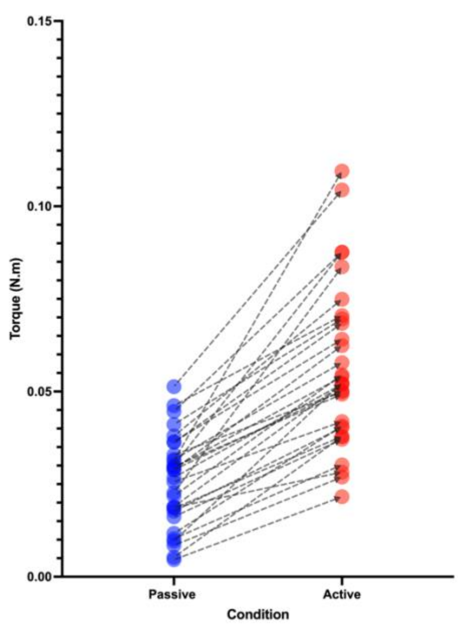


Figure 4.10 (left): Individual torque values for passive and active condition. As expected, torque values are shown to be higher for the active condition, in which participants performed an isometric contraction towards plantarflexion. A paired t-test confirmed the difference between the conditions to be significant ($p < 0.001$).

Figure 4.11 (right): A comparison between torque during the passive and active conditions. The dashed arrows indicate that all participants had a higher torque during the active condition.



Torque variability was also examined to determine how consistent participants were during each of the testing conditions. Torque variability represents participants' performance precision, where smaller values indicate better precision in performance. For example, when examining torque variability data during the active condition, we may observe how well a participant maintained their isometric contraction throughout the testing period. In other words, was the force they exerted continuous and steady or was it sporadic and fluctuating throughout testing? As for the passive condition, torque variability is reflective of the participants' ability to maintain relaxed lower limb muscles while detection threshold was being tested.

Torque variability data is presented in Figure 4.12: As expected, values appear larger and more variable for the active condition, most likely accounting for differences in participants effort/performance while sustaining the isometric contraction. A Wilcoxon Signed-Ranks test indicated that torque variability was significantly higher during the active condition ($M= 0.003 \pm 0.002$) compared to the passive condition ($M= 0.001 \pm 0.001$): ($z= -4.66$, $p<0.001$).

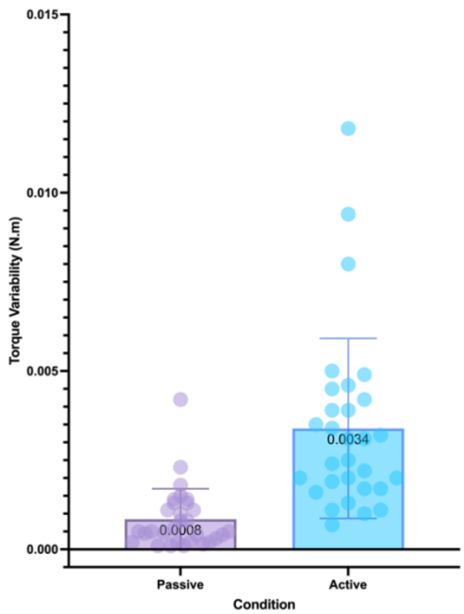


Figure 4.12: Individual torque variability values for passive and active conditions. As can be seen from the spread of data, torque variability was more widely spread during the active condition, and was confirmed to be significantly higher than torque variability during the passive condition (p<0.0001).

c. Relationship between detection threshold and torque:

Initially, we attempted to understand the relationship between mean torque and detection threshold scores in our experiment by analysing the *difference* in detection threshold in relation to *change* in torque between the passive and active conditions. However, no correlation was found ($p > 0.05$). This indicates that the change in torque likely does not influence the difference in detection threshold values between conditions. We then decided to explore the standard deviation (SD) of torque (torque variability) for each participant and its relationship with their threshold performance.

Detection threshold and torque variability:

We found significant positive correlations between detection threshold and torque variability in general and also during each of the testing conditions separately: When looking at threshold data and torque variability in general, we found a moderate positive correlation: ($r_{(29)}$= 0.47, *p< 0.001*)(Figure 4.13, black regression line). This seems to indicate that the steadier a participant was in their lower limbs in general, the better they tended to perform in terms of proprioceptive threshold. When separating data based on testing condition, we found similar results: For the passive condition (Figure 4.13, blue markers and line), we found a moderate positive correlation ($r_{(29)}$= 0.47, *p= 0.01*) and for the active condition (Figure 4.13, orange markers and line) a moderate positive correlation was also found ($r_{(29)}$= 0.44, *p=0.01*).

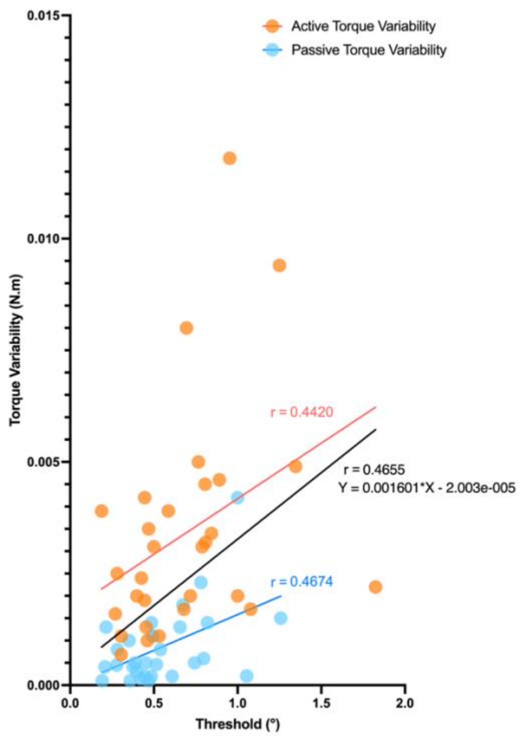


Figure 4.13: The relationship between detection threshold and torque variability. A significant correlation was found between detection threshold scores and torque variability (in black) when data for both testing conditions were taken together ($p < 0.001$), as well as when conditions were separated: Both passive (in blue, $p = 0.01$) and active (in orange, $p = 0.01$) conditions showed a significant correlation between threshold and passive and active torque variability.

Similar to the first analysis we performed between mean torque and detection threshold scores, we observed the change in torque variability in relation to the difference in detection threshold between the passive and active conditions to confirm whether participants with more *similar* torque precision between conditions also show more similar threshold scores for both conditions and vice versa. However, no significant relationships were found between the two (*p= 0.41*).

d. <u>Effect of start condition on detection threshold values</u>:

To check for any effect of "start condition" (the testing condition participants started with—either passive or active—to initiate their trials) on detection threshold scores, participants were divided into two groups based on start condition (n = 15 for each). Using a repeated measures analysis, we looked for any significant interactions between threshold scores (passive/active) and start condition (also passive/active). The results indicated there were no significant interactions within subjects (*$p = 0.578$*).

e. Effect of testing condition on frequency of incorrect judgements during testing:

During testing, a participant could answer either "up" or "down" to report the direction they perceived the foot plate was moving towards. These answers were recorded on an excel score sheet as follows:

- **Up-correct (uc):** Footplate moved upwards and participant answered correctly "up"
- **Up-false (uf):** Footplate moved upwards and participant answered incorrectly "down"
- **Down-correct (dc):** Footplate moved downwards and participant answered correctly "down"
- **Down-false (df):** Footplate moved downwards and participant answered incorrectly "up"

The frequency and ratio of uf to df scores (incorrect answers) were analysed to test for any trends in the data. Repeated measures analyses were carried out to test for any interactions between condition (passive/active) and footplate direction during incorrect answers (uf,df). However, no significant interactions were found, indicating that condition did not significantly influence the frequency of false answers in any particular direction ($p > 0.05$).

4.5 Discussion

Two complimentary studies were performed using our developed TTDPM method: The first was a preliminary study, the aim of which was to establish the normal detection threshold values for the three velocities chosen (0.1, 0.25 and 0.5°/sec) and to confirm that our method produced plausible results. The second study was our experimental study, in which we used a fixed velocity (0.1°/s) for the entire experiment. This selection was based on the range of detection threshold scores obtained in the preliminary study, as the chosen (lowest) velocity showed the highest range of scores among the three velocities, making it more likely to reveal differences between participants during our experimental trials.

4.5.1 Detection Threshold scores

For our preliminary study, average detection threshold scores of 0.76°, 0.35° and 0.18° for our chosen velocities—0.1, 0.25 and 0.5°/s respectively—indicated that increasing velocity had an inverse effect on detection threshold scores, meaning that smaller displacements were detected as the velocity of movement increased. This is in agreement with the literature (Han et al., 2016; Proske & Gandevia, 2012; Refshauge & Fitzpatrick, 1995).

Interestingly, we found the average detection threshold obtained from our preliminary study (0.76° ± 0.71) to be higher than that obtained during the passive condition in our experimental study (0.53° ± 0.25°) for the same 0.1°/s velocity, even though the method for these two tests was essentially identical (both done in a standing position while the participants legs were relaxed). In addition, the standard deviation was almost three times the size for the preliminary study. However, sample size for the experimental study was three times that of the preliminary study, which suggests the former study results to be closer to that of the general population, due to greater statistical power (Serdar et al., 2021).

As for the active detection threshold from our experimental study, average scores were found to be higher (0.69° ± 0.37°) than scores for the passive detection threshold, echoing the results found in previous studies (Kwon et al., 2013; Niessen et al., 2009; Peters et al., 2017). However, for our experiment, this difference was found to be non-significant ($p = .075$). As mentioned previously, our post-hoc analysis revealed that a sample size of at least 35 subjects were required to detect a large effect size ($d = 0.5$) with 80% power, so we presume sample size ($n=30$) was one of our limitations in this case.

4.5.2 Torque and torque variability

Recording and examining torque data in many cases serves the purpose of confirming the experiment proceeded as planned. In our case, our main objective for recording this data was to verify that participants were relaxed or contracting when they were meant to do so. After analysing several different time windows of torque data, the three second window before the footplate movement commenced was chosen to perform our analyses as it appeared to be the trace of data least affected by footplate motion.

Results indicated that mean torque values for the active condition were higher for all participants (Figure 4.10) and were found to be higher than torque values recorded during the passive condition ($p < 0.001$). This confirms that participants exerted significantly larger force through their legs during the active condition compared to the passive condition, or in other words, were performing the experiment correctly. Torque variability data was also analysed during the passive and active conditions, to examine participants' consistency in maintaining either relaxed legs during the passive condition, or an isometric contraction during the active condition. Similar to our findings with mean torque, torque variability was significantly higher during the active condition ($p<0.001$). As torque variability here demonstrates imperfection in the task of holding a constant torque, these results indicate that participants were able to maintain a significantly steadier torque during the passive condition, when they were not required to hold a steady contraction, than during the active condition.

We then went on to analyse the relationship between torque and detection threshold: We looked at the change in baseline torque values between conditions and whether this had any influence on the difference in threshold scores between passive and active. However, results indicated there was no significant effect between the two. When referring back to our experimental method, it may be determined that torque values were "controlled" within the specific range practiced before the trial (between 5-15% RVC) and monitored for throughout the duration of the experiment. This indicates that analysing torque against threshold scores may be slightly confounded, as participants were limited to a certain percentage of force with which they were allowed to use during testing (10% of RVC). In addition, torque was monitored throughout testing to confirm participants adhered to this range of force. Torque variability, on the other hand, may indicate a true, uncontrolled effect, as it demonstrates participants' ability to maintain either a steady contraction during the active phase, or a relaxed stance during the passive phase. To that end, torque variability was analysed against detection threshold scores as a whole, and during each of the experimental conditions separately (passive and active), and significant positive correlations were found for all the aforementioned pairings: Moderate positive correlations were found between torque variability and detection threshold in general ($p<0.001$), and between torque variability and detection threshold during both passive and active conditions (both $p<0.01$). These results seem to imply that the steadier or more stable a participant was in their lower limbs throughout the trials, the better their ankle proprioceptive acuity appeared to be.

However, when we analysed the change in torque variability between the passive and active conditions in relation to the *difference* in detection threshold between the passive and active conditions, no significant effect was found. Taken together with our previous findings—that smaller torque variability was associated with lower threshold scores, and larger torque variability was associated with higher threshold during both conditions in a linear correlation—this was a surprising discovery. It was expected that participants with *similar* torque precision between conditions—meaning they either maintained torque well in both passive and active conditions, or conversely, did not maintain torque very well during both conditions (difference here would also be small)—would also show *similar* threshold scores (smaller difference) between passive and active conditions. However, this was not the case. This may suggest that participants with similar torque variability between conditions have very *dis*similar scores between the passive and active condition. And so we conservatively suggest that muscle contraction may have indeed exerted an effect on proprioceptive acuity in this case; however, for our study this effect did not reach significance.

4.5.3 Relevance of findings and clinical implications

Steadiness, or stability, may be defined as showing little variation or fluctuation (Merriam-Webster, n.d.). In our experiment, torque variability was a measure of participants stability, as it demonstrated how well they were able to maintain force (or lack thereof) without any fluctuations, or sudden changes, in this force.

Although no studies on the effect of steadiness on proprioception could be found to directly compare our results with, our findings reflect the idea that in the triceps surae, spindle sensitivity is enhanced mostly when the muscle is relaxed. This is in agreement with Peters et al. (2017) in that muscle spindle sensitivity is increased with decreased muscle activation. However, this concept applies to results from our passive condition only. During the active condition, participants who were best able to maintain an active isometric muscle contraction in their calf muscle—or in other words, had the most stable calf-muscle activity—also achieved the best detection threshold scores. This also implies that participants who were worst at maintaining a stable isometric muscle contraction in their calf muscle achieved the worst detection threshold scores. It may be that the high torque variability these participants demonstrated, served as additional background noise throughout the muscle, so that it became harder to distinguish externally-imposed changes in muscle length (footplate movement) from self-generated changes within the muscle (isometric contraction). Therefore, it may be reasonable to conclude that proprioceptive acuity of the ankle joint is enhanced when background noise within the muscles surrounding the joint is reduced.

And so the question that comes to mind is: How do we apply this concept to clinical populations that suffer from proprioceptive deficits? A possible answer is by focusing on isometric muscle work during rehabilitation, in order to improve patients' ability to maintain stable muscle contractions for increased lengths of time. According to our findings, this may enhance proprioceptive acuity, which will, in turn, improve motor control during activities of daily living.

4.5.4 Limitations

As mentioned previously, a limitation in our study was sample size, as post-hoc analysis indicated a sample size of at least 35 subjects was needed to detect our effect size with 80% power.

Another limitation of our study was testing time: This varied greatly from participant to participant, from 45 minutes to well over 75 minutes in standing. Although comfort breaks were taken throughout testing, it was apparent that this method would be difficult to implement with older subjects, and we concluded it was unfortunately impractical for that means. Therefore testing was limited to young healthy participants.

4.6 Conclusion

Our study aimed to test the effect of isometric contraction on proprioceptive acuity (via detection threshold) at the ankle. Participants were tested under two conditions: in the first condition they held an isometric contraction in the calf muscle while performing our detection threshold task; in the second, they performed the same task while the calf muscle was as relaxed as possible. A torque trace was recorded throughout the trials to ensure participants' muscles were contracted or relaxed when required.

Although our data showed that proprioceptive acuity was not significantly affected by muscle contraction, we found some interesting trends in our torque data. In particular, we found a significant positive correlation between torque variability and threshold scores for both the passive and active conditions. Our results seem to suggest that the steadier a participant is, the better they can detect footplate movement.

Chapter 5: Proprioceptive Reaction Time as a Predictor of Body Sway

Abstract

For our last experiment, we aimed to determine whether proprioceptive acuity at the ankle joint, measured through reaction time threshold, could predict quiet standing balance. A total of 61 healthy adults, aged between 19-90 years, and divided into a young (21.31 ± 3.83 years) and old (75.15 ± 3.89 years) age group, participated in the study. Our second TTDPM method, reaction time threshold, was used to assess proprioceptive acuity at the ankle, in which detection threshold is measured in time (seconds (s)). In addition, quiet balance performance was assessed using a force platform to measure participants' body sway during four different quiet standing conditions: (1) eyes open-feet apart (EOFA), (2) eyes open-feet-together (EOFT), (3) eyes closed feet-apart (ECFA) and (4) eyes closed feet-together (ECFT). Our main outcome measures were reaction time (s), and centre of pressure (COP) velocity measured in the anterior-posterior (AP) and mediolateral (ML) planes. Our results indicated that age significantly predicted reaction time (z=-5.08, $p<0.001$). In addition, we found a significant interaction between age and vision for all body sway outcome measures (COP-AP and COP-ML velocity and COP speed)(all $p<0.05$). This indicates that older participants rely on vision more than younger participants to maintain their standing balance. Lastly, we found that reaction time significantly predicted COP AP body sway performance when feet were apart. This includes both EOFA and ECFA conditions. Our findings may have direct clinical implications, as

they indicate a clear link between proprioceptive and balance performance, which may be utilised as an evidence-based method for balance rehabilitation of clinical populations.

5.1 Introduction:

In psychophysics, the magnitude of a stimulus that will lead to its detection 50% of the time can be defined as a threshold. It may also be described as the minimum intensity of a stimulus that is required to elicit a response and may be used to test a variety of stimuli, including auditory, visual and, as we demonstrated in Chapter 4, proprioceptive stimuli. A significant limitation of our previous experiment was that the method we used to establish the threshold could be very time consuming, and this was decided impracticable for older aged participants. Using data from our active-passive threshold experiment, we were able to develop a much simpler testing method: Instead of using amplitude (°) as our outcome measure for threshold, we found it to be much more straightforward to record participants' reaction time (milliseconds (ms)) as a measure of proprioceptive acuity at the ankle. Technically, **reaction time** would allow us to establish detection threshold using a more concise testing method, which would be more suitable for testing our elderly population, especially since we planned to test it along with **quiet standing balance** via body sway, a separate test in itself, to determine whether a relationship exists between the two outcome measures.

5.1.1 Literature review

Reaction time (RTm) may be defined as the time elapsed between the application of a stimulus and the indication of perception by the subject to whom the stimulus was applied (Reaction time-APA Dictionary of Psychology, n.d.). Essentially a sensorimotor process, reaction time involves the receipt of sensory information, the central processing and integration of this information, and finally the execution of a motor response (Balakrishnan et al., 2014). However, as indicated in our definition, it is always a voluntary process, requiring “perception” of the stimulus.

The use of reaction time in the literature in the context of proprioception is quite limited, as it is more well-known as a measure of performance level in sport (Nuri et al., 2013; Badau et al., 2018; Sainz et al., 2020) or as an indicator of motor ability alongside muscle strength and balance for example, in healthy or clinical populations (Lord et al., 1993; Tsang and Hui-Chan, 2003; Mazbouh et al., 2023). However, technically speaking, reaction time may be precisely interpreted as the minimal time necessary to process proprioceptive information and produce a voluntary motor response (Verschueren et al., 2002), making it seemingly well-suited as an outcome measure for proprioceptive threshold detection.

Among the more recent literature on ankle proprioception, only two studies were found which use reaction time to quantify proprioceptive acuity. The first by Verschueren et al. (2002) used reaction time threshold (among other tests) to assess whether ankle proprioception declines as a result of ageing. A total of 47 elderly subjects and 15 young subjects participated in the study. Participants were seated with their lower legs secured so that movement was restricted at the ankle joint. The ankle was then passively rotated by a torque motor and a potentiometer transduced the ankle angle. A total of 10 trials were recorded for each participant. Movement sensation at the ankle was assessed by asking participants to open their hand as soon as they felt their ankles move. Hand opening was detected by breaking the electric continuity through metal contacts circling the thumb and index finger of the right hand. Although results showed that reaction time was not significantly different between young adults and the elderly ($p > 0.05$), the velocity chosen (30°/s) may have been too high to capture a difference between these age groups, as it was found in our own preliminary experiment (see Chapter 4: 4.4.1, Figure 4.4) that the higher the velocity chosen, the smaller the RTm variability tended to be between participants.

The second study using reaction time to measure proprioceptive acuity was by Djajadikarta et al. (2020) who investigated whether ankle proprioception changes with age in healthy, community-dwelling people. The study included 80 participants aged 19-80 years. Participants sat with their leg extended while their foot rested on a motorised footplate. Thixotropic changes were controlled for by repetitively and passively moving the ankle through a set pattern of motion before each test trial. The motorised footplate moved at a speed of 13°/s through the dorsi/plantarflexion plane, and a table was positioned over the legs to occlude vision of the legs and feet. Proprioceptive reaction time was measured as a plantar-flexor response to a suprathreshold movement of the footplate. Similar to our experiment, data from a previous detection threshold experiment was used to set the footplate amplitude for the reaction time experiment. Footplate amplitude was set to twice the detection threshold found in the previous test to ensure participants could detect the movement. Participants were instructed to push down with the ball of their foot (i.e., plantarflex) as soon as they felt the footplate move, and the footplate moved randomly towards both plantarflexion and dorsiflexion. Mean reaction time was found to be 0.251 ± 0.054 s. However, there was no relationship between age and proprioceptive acuity (*p=0.58*). Again in this study, the velocity chosen seems high (although not nearly as high as the study before it) and if our previous detection threshold experiment is any indication, threshold scores tend to become less variable between participants the higher the velocity. In our case (see Chapter 4) the velocities chosen (0.1, 0.25, 0.5°/s) were but a fraction of the velocities used in the studies by Verschueren et al. (2002) and Djajadikarta et al. (2020). However, we still saw this inverse effect in our study, with the slowest velocity (0.1°/s) exhibiting the largest range in threshold scores, and the fastest velocity (0.5°/s) showing the smallest range in threshold scores among the three velocities. Although it seems likely, it is yet to be determined whether we will see the same effect with reaction time scores. The seemingly reciprocal relationship between ankle proprioception and body sway in the literature was one of the reasons body sway was chosen as an outcome measure in our study. As mentioned previously in our introduction, it is thought that receptors in the ankle region provide the most important information with regard to body sway during quiet standing, contributing approximately 58% to body sway during standing (Lord et al., 1991; Henry and Baudry, 2019). In addition, several studies have observed a noticeable increase in body sway when proprioceptive signals are disrupted during standing (Inglis et al., 1994; Lajoie et al., 1996; Horak et al., 2009). Moreover, muscles surrounding the ankle, dorsi- and plantarflexion musculature in particular, have been shown to greatly influence body sway COP scores (Warnica et al., 2014; Yoshizawa and Yoshida, 2022).

Taken together, the literature seems to indicate a good chance of finding a correlation between ankle proprioception and body sway during quiet standing.

With regard to body sway in our experiment, our initial plan was to compare it with different types of proprioceptive tests to establish which best predicted body sway performance. Due to our late realisation that the previous detection threshold experiment was unfit for older participants, as well as several time constraints which hindered our ability to continue work in the lab, including covid as well as personal extenuating circumstances, we were only able to make this comparison between body sway and reaction time threshold.

5.1.2 Aim of study

Our study aimed to test whether proprioceptive acuity at the ankle joint correlated with body sway during quiet standing. We analysed this relation in healthy young, middle-aged and elderly participants. To do this, proprioceptive acuity was measured using a reaction time method developed specifically for our lab setup (Chapter 2) and body sway was measured using a force plate. Although this experiment required two separate tests, it took no more than 35-40 minutes to complete both for participants of all ages. For the reaction time threshold test, we measured the time (s) it took participants to detect movement of the footplate, whereas for the body sway test we measured centre of pressure (COP) velocity (mm/s) during four different conditions of quiet standing.

5.1.3 Hypotheses

We hypothesised that reaction time to imposed ankle joint movement would significantly correlate with body sway.

Null Hypothesis (H_0):

-Proprioceptive acuity at the ankle as measured by reaction time threshold will have no significant relationship with quiet standing balance as measured by bipedal body sway.

Alternative Hypothesis (H_A):

-Proprioceptive acuity at the ankle as measured by reaction time threshold will have a significant relationship with quiet standing balance as measured by bipedal body sway.

5.2 Methods

Similar to our previous threshold study, we conducted a separate preliminary study before starting our experimental trials to investigate the average ranges of reaction time threshold of ankle proprioception in the population using different velocities. The velocities chosen for reaction time for our preliminary study were 0.25, 0.5 and 0.75°/s. Whereas for our experimental study, either 0.25 and 0.75°/s, **or** 0.5 and 0.75°/s were chosen to conserve time as using faster movements and/or less conditions would be more practical for our elderly participants. Throughout this chapter, sections will be divided to include both preliminary and experimental studies.

5.2.1 Participants

For the preliminary study, ten healthy young adults (5 male) with a mean age of 23.8 ± 5.37 years were recruited via email between May and June 2021 to participate in the study. As for the experimental study, it was carried out on two different age groups at two different time periods. Thirty five healthy young adults (21 male) with a mean age of 21.31 ± 3.83 years were recruited via email and participated in our study between September and November 2021, whereas twenty six healthy older participants with a mean age of 75.15 ± 3.89 years were recruited—from the '1000 elders' volunteer list compiled by Professor Janet Lord—and participated in our study between March and May 2022. Inclusion criteria for both experiments were the same as for our previous experiments: Healthy without any ongoing physical or neurological injury or illness which might affect normal control of standing balance and/or any injury to the lower limbs. The experiments took place in the Human Balance Laboratory in the School of Sports, Exercise and Rehabilitation in the University of Birmingham, and each took approximately 35-40 minutes to complete per participant. The experiment was approved by the School of Sport, Exercise and Rehabilitation Ethical Review Committee at the University of Birmingham (ERN_14-0740A) and performed in accordance with the Declaration of Helsinki (World Medical Association, 2013). Participants were given information sheets on the experiments and provided written, informed consent before starting the trials.

5.2.2 Experimental setup

To assess reaction time (RTm) for both our preliminary and experimental studies, we used the same equipment and positioning as we did for our previous experiments, and participants stood barefoot with their back leaning against a board angled 21.8° from vertical (Figure 5.1). A Velcro strap was fastened around the lower thighs to minimise any movement of the knees. A foam cushion was placed behind both knees for comfort. Again, this setup was chosen to ensure the participant was bilaterally weight bearing without needing to balance themselves and allowed them to maintain standing with minimal muscle activity. Each foot was placed on a separate motorised footplate, which could be rotated individually using two linear motors attached by levers (Model XTA3810S; Copey Motion Systems LLC, Basildon, UK). The axis of rotation of each footplate was made to be approximately collinear with the ankle joint. In addition, each foot plate was affixed with sensors for position (Model CP-2UT; Midori Precisions Co., Tokyo, Japan) and torque (Model 31, Sensotec Inc., Columbus, OH, USA).

For the experimental study, an additional assessment was carried out after completion of the RTm assessment. Body sway (BSw) was assessed during quiet bipedal stance using a Kistler force plate (Kistler type 9260AA6). Participants were asked to stand quietly on the Kistler force plate, facing a plain board which stood at a 90 cm distance away from the centre of the force plate. Markings placed vertically (at midline) and horizontally (20 cm distance between big toes) on the force plate allowed participants to position their feet in the same position consistently for consecutive trials.

5.2.3 Procedure

Preliminary study

To assess RTm, footplate movements were generated through Matlab (Mathworks Inc., Natick, MA, USA) to move the **right** footplate either up (dorsiflexion) or down (plantarflexion) from the start position of 0^o (perpendicular to angled backboard). The **left** footplate remained at the reference 0^o start position during the entire experiment. Participants were given a button press (Figure 5.1) to hold in their dominant hand and were asked to press it as soon as they perceived ankle motion, then call out the direction of motion: either “up” or “down”. These answers were recorded in an excel sheet to determine how often participants correctly perceived the passive ankle movement direction during testing.

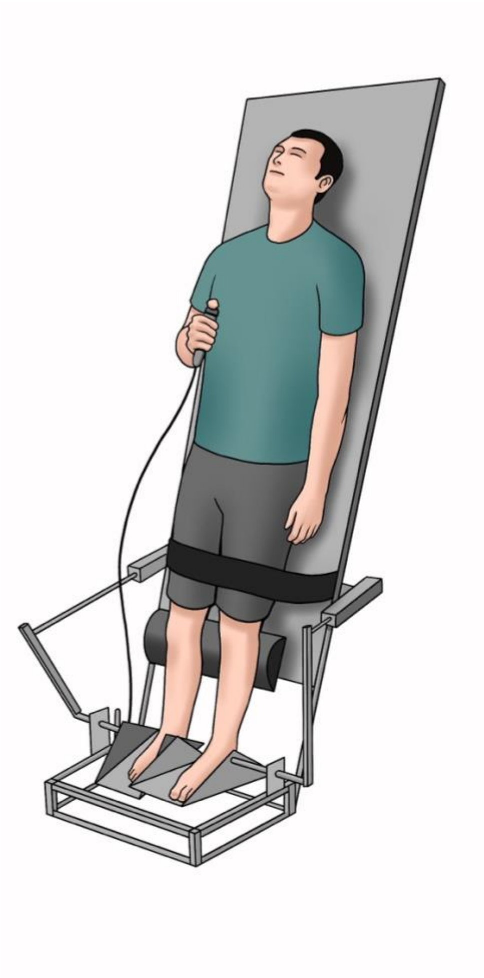


Figure 5.1: Participant positioning while performing the reaction time task. With a button press held in their dominant hand, participants were asked to keep their eyes closed and their body completely relaxed during

Proprioceptive RTm was measured as a thumb-flexor response to a suprathreshold movement of the footplate. The size of the footplate movement (°) was set to more than seven times the detection threshold recorded during our previous threshold experiment. This was to ensure that participants would be able to detect footplate movement within the time frame allotted for each velocity (see Figure 5.2a-b and c).

For the preliminary study, only RTm was assessed and a total of nine trials (three for each of the following velocities: 0.25°/s, 0.5°/s and 0.75°/s) were tested per participant. Each trial consisted of ten random movements, half of them upward (five dorsiflexion rotations) and the other half downward (five plantarflexion rotations) (Figure 5.2). This adds up to 30 movements per velocity for each participant, meaning they were allowed to judge the direction of ankle rotation a total of 30 times for each velocity. For each trial, any RTm value that occurred so late that the platform already changed direction (see Figure 5.2 for clarification) was removed from the final data i.e. > 10 s for the 0.25°/s condition, > 5 s for the 0.5°/s condition and > 3 s for the 0.75°/s condition as they were considered "fails". Additionally, any trials where subjects reported the wrong direction were also considered fails and removed from final data. Furthermore, if participants were unable to correctly perceive direction of ankle movement (or were unable to perceive footplate movement in time) for at least 7 out of the 10 iterations, that trial would not be included in the final data. This was in line with the "70% detection level" method adopted by Refshauge and Fitzpatrick (1995) for their detection threshold study.

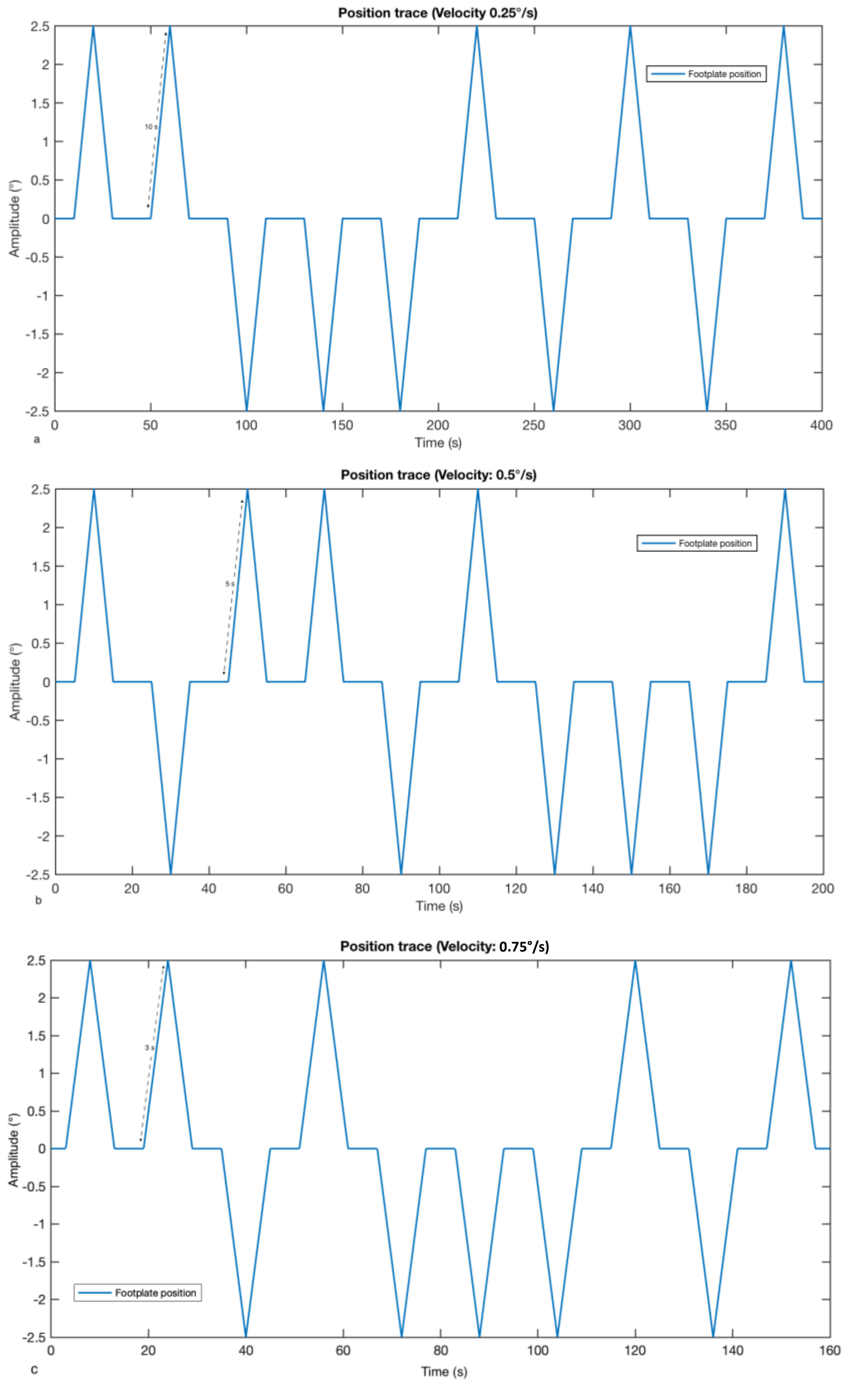


Figure 5.2: Position traces showing footplate movement during three different reaction time trials. (from the top) Velocity conditions were 0.25, 0.5, and 0.75°/s)**.** Each of the 3 trials shows a random sequence of ten footplate movements, moving either upwards toward dorsiflexion or downwards toward plantarflexion. As shown on the x-axis of each of the graphs, trial time ranged from 160-400 seconds, which equates to 2.6 to 6.6 minutes per trial, depending on the velocity chosen. For all of the trials, the footplate moved a total of 2.5°, which is more than 7 times the detection threshold recorded in our previous threshold experiment (0.35° ± 0.35 for 0.25°/s). However, due to the difference in velocity for each of the three trials shown, the time it took the footplate to reach 2.5° and subsequently change direction differed: For the slowest condition (0.25°/s) this took 10 s (5.2a), for the medium condition (0.5°/s) it took 5 s (5.2b), and for the fastest condition (0.75°/s) it took 3 s (5.2c) for the footplate to reach maximum amplitude (2.5°) before it changed direction.

After completion of all trials, we recorded the median value of each participants' RTm scores rather than the mean value, as the median tends to be less swayed by outliers for RTm assessments. So each participant ended up with three RTm scores-one score for each of the tested velocities.

Experimental study:

As for the experimental study, we assessed RTm using the same set-up as was used for the preliminary study. However, only two velocities per participant were tested this time around: the velocities which showed more variability in our preliminary study (0.25°/s and 0.75°/s) were used to test the young age group (n=35), whereas the two faster velocities (0.5°/s and 0.75°/s) were used for the older aged group (n=26) to save time and reduce participants' fatigue as much as possible.

As was the case in the preliminary Rtm study, participants were given a button to hold in their dominant hand and were asked to press it as soon as they perceived ankle motion, then immediately calling out the direction of motion: either "up" or "down" (Figure 5.1). These answers were recorded in an excel sheet to determine how often participants correctly perceived passive ankle movement direction during testing. A total of six trials (three per velocity, in random order) were tested per participant. Identical to the preliminary study, each trial consisted of ten random movements, half of them upward (five dorsiflexion rotations) and the other half downward (five plantarflexion rotations) (Figure 5.2). This adds up to 30 movements per velocity for each participant, meaning they were allowed to judge the direction of ankle rotation a total of 30 times for each velocity. Again, for each trial, if participants were unable to correctly perceive ankle movement direction for at least 7 out of the 10 iterations, that trial would not be included in the final data.

Following the RTm assessment, we assessed BSw during quiet bipedal stance, specifically Centre of Pressure (COP) velocity (separately in the anterior-posterior and mediolateral direction) and speed of body sway, using the Kistler force plate. COP velocity represents the total distance travelled in a specific direction by the COP over time, and has been shown to be a reliable measure when a double-legged stance is used (Palmieri et al., 2002). It is thought that an increase in COP velocity represents a decreased ability to control posture, whereas a decrease in velocity represents a greater ability to maintain balance in standing (Baloh et al., 1998a, 1998b; Tjon A Hen et al., 2000). As for COP speed, it corresponds to the cumulative distance travelled by the COP over time, while negating the direction of movement. This value was obtained from participants' velocity measurements using a simple mathematical equation on Matlab.

For the BSw part of the experiment, participants' balance was assessed during four different conditions of quiet standing:

1) Eyes open with feet apart (EOFA)
2) Eyes open with feet together (EOFT)

3) Eyes closed with feet apart (ECFA) or
4) Eyes closed with feet together (ECFT)

For the feet apart conditions, participants placed their feet so that their big toes were separated 20 cm from each other, and for the feet together conditions, participants stood with their feet touching at midpoint of the force plate. Markings on the force plate ensured foot placement was consistent for all trials. Each of the four quiet standing conditions (EOFA, EOFT, ECFA and ECFT) were tested three times in random order (for 40 seconds each) for a total of 12 trials recorded. Participants were asked to relax and maintain quiet standing for the duration of testing. For each condition, anterior-posterior velocity (APV), mediolateral velocity (MLV), and overall speed data was gathered. The average score of the three trials recorded was calculated so that each participant ended up with an AP velocity score, an ML velocity score, and an overall speed score for each of the four conditions.

5.3 Analyses

Graph pad (Prism) and SPSS software were used to analyse all data. Initially, descriptive statistics were performed on the data to get an overall picture of our results.

Data was then tested for normal distribution using the Shapiro-Wilk test to determine the appropriate inferential tests to be applied to the data.

For the preliminary study, data for all three tested velocities were found to be normally distributed ($p>0.05$), therefore parametric tests were used for our inferential analyses. A one-way repeated measures analysis of variance (ANOVA) with Geisser-Greenhouse correction was used to determine if there was a difference between RTm scores for the three chosen velocities. It was also used to check whether direction of ankle rotation affected RTm scores during our preliminary study.

As for the experimental study, for the young age group, we found that Rtm was non-normally distributed ($p<0.0001$), whereas the old age group RTm data was normal ($p=0.26$). However, pooled together, RTm data was ultimately non-normally distributed. As for the BSw data, it was found to be non-normally distributed ($p < 0.0001$), therefore non-parametric inferential tests were used for analyses of all the experimental study data, except for when we were analysing the old age group RTm data separately. A Wilcoxon Signed-Ranks test—the non-parametric equivalent of a paired t-test—was used to compare RTm scores between velocities for our young age group, while a paired t-test was used for our older aged group. In addition, a Mann-Whitney U test—the non-parametric equivalent of an independent t-test—was used to assess whether RTm was different between the two age groups. We also used a Friedman test—the non-parametric equivalent of a repeated measures ANOVA—to detect differences within our BSw data, which was followed by multiple comparison tests on the data using the Wilcoxon Signed-Ranks test. A Kruskal-Wallis test—the non-parametric equivalent of a one-way ANOVA—was used to define any differences between young and old participants with regard to BSw.

Lastly, to explore any potential correlations between RTm and BSw data in our experimental study, the Spearman correlation test—the non-parametric equivalent of a Pearson correlation—was used. Additionally for the young age group, a Wilcoxon Signed-Ranks test was used to examine any differences between dorsiflexion and plantarflexion Rtm for each of the two velocities (0.25°/s and 0.75°/s).

5.4 Results

5.4.1 Preliminary study

Data from the preliminary RTm experiment will be presented under the following subcategories:

1. Demographic data
2. Descriptive statistics
3. Inferential statistics

1. Demographic data:

A total of 10 healthy young adults (5 males, 5 females) with a mean age of 23.8 ± 5.37 years participated in the study. Participants' average height was 171.1 ± 14.99 cm, average weight was 65.89 ± 16.05 kg, and average BMI was 22.27 ± 2.98 kg/m^2.

2. Descriptive statistics:

Results showed that on average RTm decreased with the increase in angular velocity (Figure 5.3).

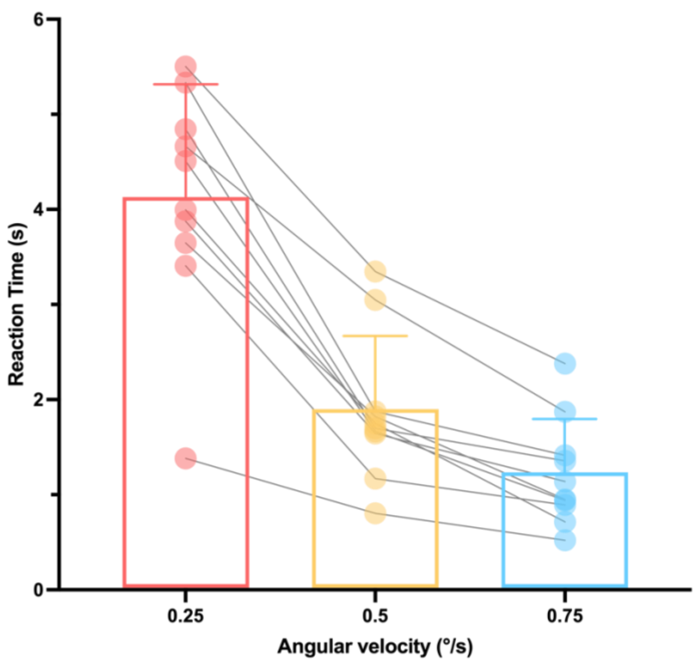


Figure 5.3: Individual reaction time scores during our preliminary study. A clear inverse relationship can be seen between reaction time and angular velocity, where reaction time increases as velocity decreases. Mean and standard deviation bars shown.

As can be seen in Figure 5.3, all participants were able to detect footplate movement the fastest at the highest velocity (0.75°/s), and the slowest at the lowest velocity (0.25°/s). Mean RTm was 4.12 (± 1.19), 1.88 (± 0.77), and 1.22 (± 0.56) seconds for 0.25, 0.5, 0.75°/s velocities respectively. This was an expected result, as a similar trend was apparent in our detection threshold experiments.

3. Inferential statistics:

To check whether the difference between RTm scores was significant for the three chosen velocities, a one-way repeated measures analysis of variance (ANOVA) was performed with a Geisser-Greenhouse correction. Results confirmed that the difference was indeed significant ($F_{(1.247, 11.22)}$=92.76, *p<0.0001*) indicating an effect of velocity on RTm scores.

Data was further analysed to check whether direction of rotation had any effect on RTm scores using a general multivariate test. Figure 5.4 shows individual RTm during upward rotation/dorsiflexion (DF) and downward rotation/plantarflexion (PF) for each of the three angular velocities.

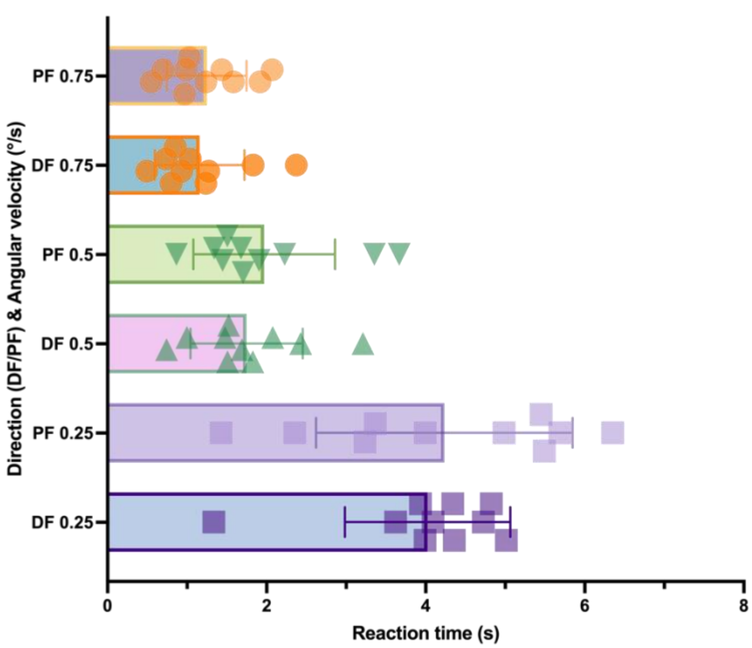


Figure 5.4: Individual reaction time scores during dorsiflexion (DF) and plantarflexion (PF). Although reaction time scores tended to be lower on average during DF for all velocities, no significant interaction was found between direction of rotation and reaction time scores for any of the three velocities (all *p>0.05*).

Our results showed that there was no interaction between direction of rotation and reaction time scores (*p>0.05*). This indicates that there was no significant difference in reaction time between DF and PF for any of the tested velocities.

5.4.2 Experimental study

Data from the preliminary RTm experiment will be presented under the following subcategories, divided as shown:

1. Demographic data
2. Descriptive and inferential statistics
    a. Reaction time
    b. Body sway
    c. Reaction time and body sway

1. Demographic data:

A total of 61 healthy young and elderly adults with a mean age of 21.31 ± 3.83 years and 75.15 ± 3.89 years respectively participated in the study. Mean age was separated for each group as there was a large age gap between them, and separating the values would be more informative rather than pooling the ages together. However, participants' pooled (1) average height was 173.2 ± 11.56 cm, (2) average weight was 70.01 ± 12.54 kg, and (3) average BMI was 23.24 ± 2.79 kg/m^2.

Separating age groups into young and older age groups, remaining demographic data was as follows:

-Young (n = 35): Average height was 176.4 ± 10.14 cm and average BMI was 23.01 ± 2.78 kg/m^2.

-Old (n = 26): Average height was 169 ± 12.18 cm and average BMI was 23.54 ± 2.84 kg/m^2.

Before analysing our data, we performed a post-hoc analyses which indicated that our sample size (n=61) should result in 99% power to detect a true correlation (r=0.5, α = 0.05).

2. Descriptive and inferential statistics:

   a. Reaction time:

   For the young age group, we assessed RTm for 0.25°/s and 0.75°/s velocities, whereas for the older age group, RTm was assessed for 0.5°/s and 0.75°/s velocities (Figures 5.5 and 5.6). This was done to save time and decrease the chances of older aged participants feeling fatigued during testing, as the assessment for the 0.25°/s was the most time-consuming of the three velocities. A Wilcoxon Signed-Ranks test indicated that regarding the young age group, the RTm for the 0.25°/s velocity was significantly higher than that for the 0.75°/s velocity (z = -5.16, *p<0.001*).

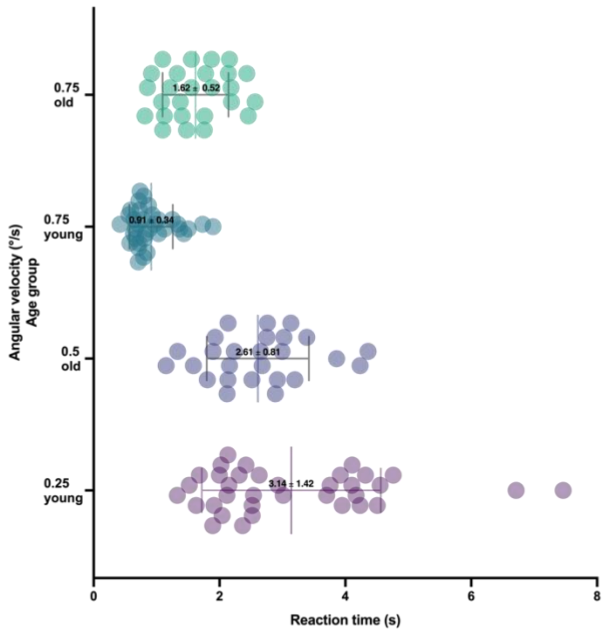


Figure 5.5: A comparison of reaction time scores between young and old age group. Individual reaction time scores are shown with mean and standard deviation bars. The scores are only directly comparable between the age groups for the fastest velocity (0.75°/s). Nonetheless, a significant difference was found between the lower and higher velocities for each the young (0.25 and 0.75°/s, *p<0.001*) and old (0.5 and 0.75°/s, *p<0.0001*) age groups.

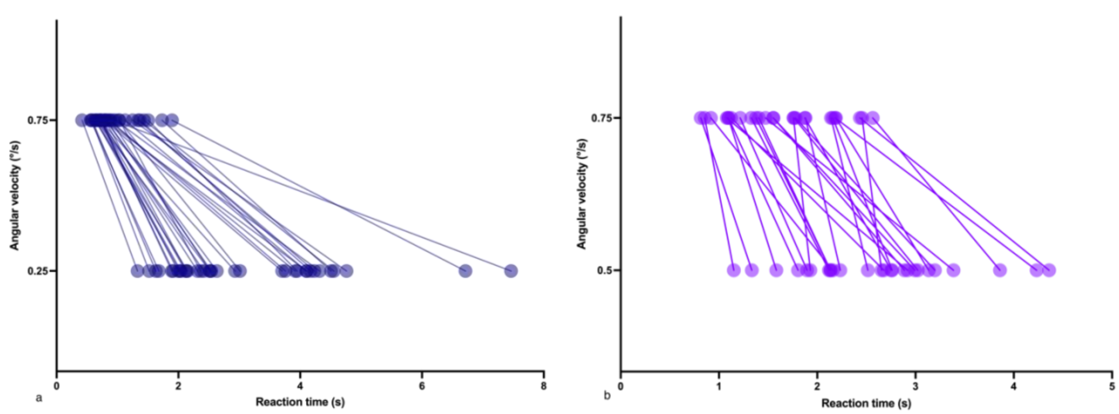


Figure 5.6: A comparison of reaction time scores within age groups. All participants of both young (a) and old (b) age groups scored lower (faster) reaction time scores for the higher of the velocities tested in each group.

For the old age group, a paired t-test revealed a significant difference between RTm scores as well (t (25) = 8.85, *p<0.001*), indicating that participants scored significantly lower (better) RTm for the faster velocity (0.75°/s) than they did for the slower velocity (0.5°/s).

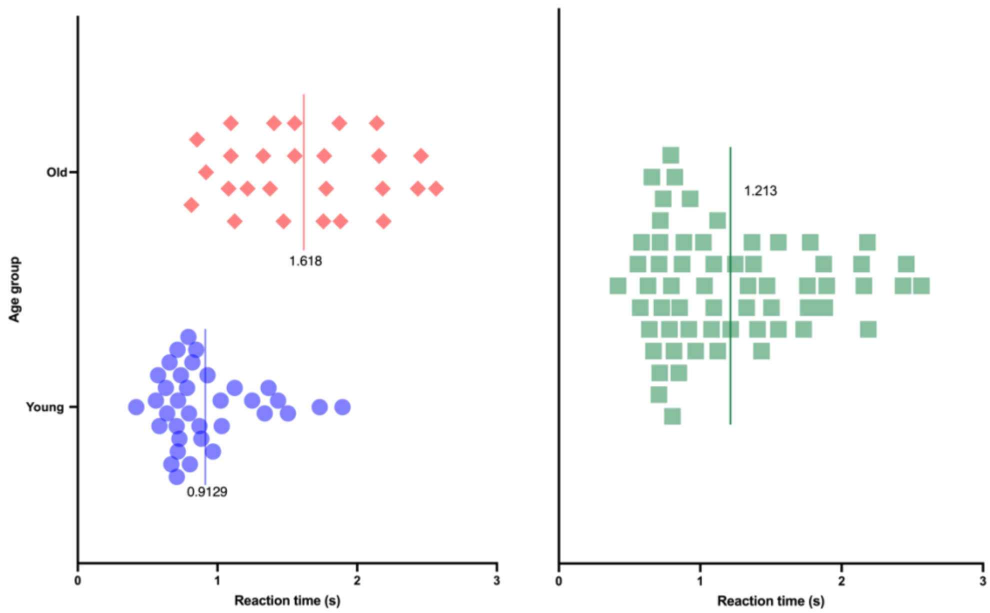


Figure 5.7: Individual RTm scores for the 0.75°/s velocity for all participants with mean value: The left graph shows the effect of age on reaction time, which was found to be significant (p<0.001), confirming that age significantly slowed proprioception as measured by reaction time. The graph on the right shows all participants' scores combined.

Figure 5.7 shows individual RTm scores (0.75°/s) for all participants, separated by age group (left) and combined (right). It is apparent from the left graph that young participants scored better RTm scores on average. To evaluate whether the difference in RTm was significant between the two age groups, a Mann-Whitney U test was performed, and confirmed that the young age group scored significantly lower RTm scores than the old age group (z=-5.08, *p<0.001*).

Differences in reaction time between dorsiflexion and plantarflexion:

For the young age group, the RTm data was further analysed to assess whether direction of ankle rotation (up or down) affected participants RTm scores. Figure 5.8 illustrates individual RTm scores separated into those taken during DF and PF for each of the two tested angular velocities (0.25°/s and 0.75°/s).

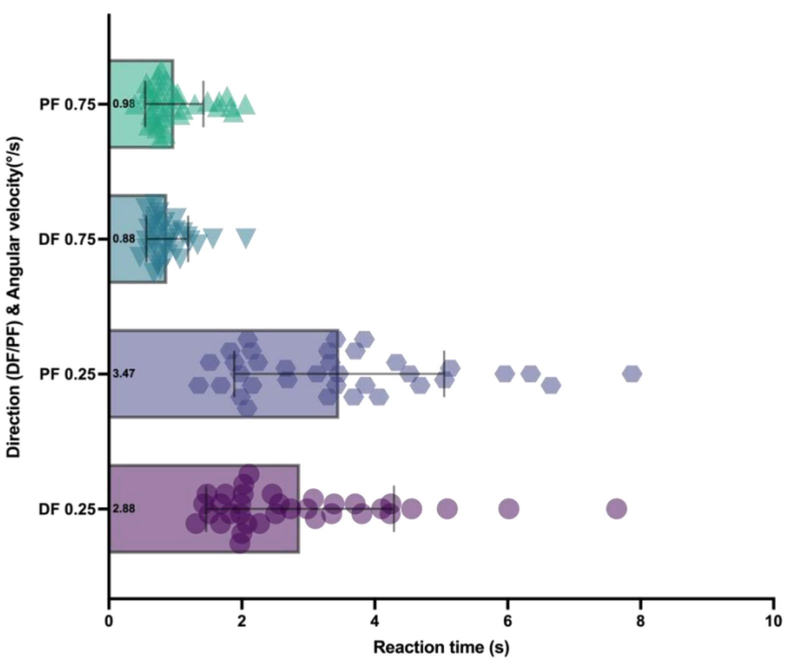


Figure 5.8: Individual reaction time scores during dorsiflexion (DF) and plantarflexion (PF) during the experimental study. Data was taken from the young age group only. Dorsiflexion reaction time was shown to be significantly lower than that of plantarflexion for both tested velocities (p < .05). Mean and standard deviation bars shown.

A Wilcoxon Signed-Ranks test indicated that dorsiflexion RTm was significantly lower than plantarflexion RTm for both 0.25°/s (z = -3.05, *p=0.002*) and 0.75°/s (z = -2.09, *p=0.037*) angular velocities. This indicates that participants perceived motion of the ankle faster when the ankle was rotated upwards. For the older participants, data for DF and PF RTm was unfortunately not available.

b. Body sway:

As mentioned previously, four different conditions of BSw were tested: EOFA, EOFT, ECFA, ECFT. For each of the conditions, data for three outcome measures based on BSw direction-AP velocity, ML velocity and speed (velocity without direction)-were recorded (mm/s). Descriptive analyses (Table 5.1) show that on average participants swayed the slowest during the EOFA condition and fastest during the ECFT condition (also see Figure 5.9).

Outcome measure /Condition	Eyes open-feet apart (EOFA)	Eyes open-feet together (EOFT)	Eyes closed-feet apart (ECFA)	Eyes closed-feet together (ECFT)

Table 5.1: Descriptive statistics of body sway data for all participants. Body sway was found to be slowest during the EOFA condition, whereas the ECFT condition showed the fastest body sway movement of the four conditions.

Anterior-posterior velocity (mm/s)	7.46 ± 2.85	8.91 ± 3.4	11.96 ± 7.74	17.91 ± 12.07
Medio-lateral velocity (mm/s)	4.58 ± 1.88	10.5 ± 3.92	5.33 ± 2.33	19.83 ± 11.37
Speed (mm/s)	7.31 ± 2.5	11.66 ± 4.14	10.81 ± 6.3	22.71 ± 13.86

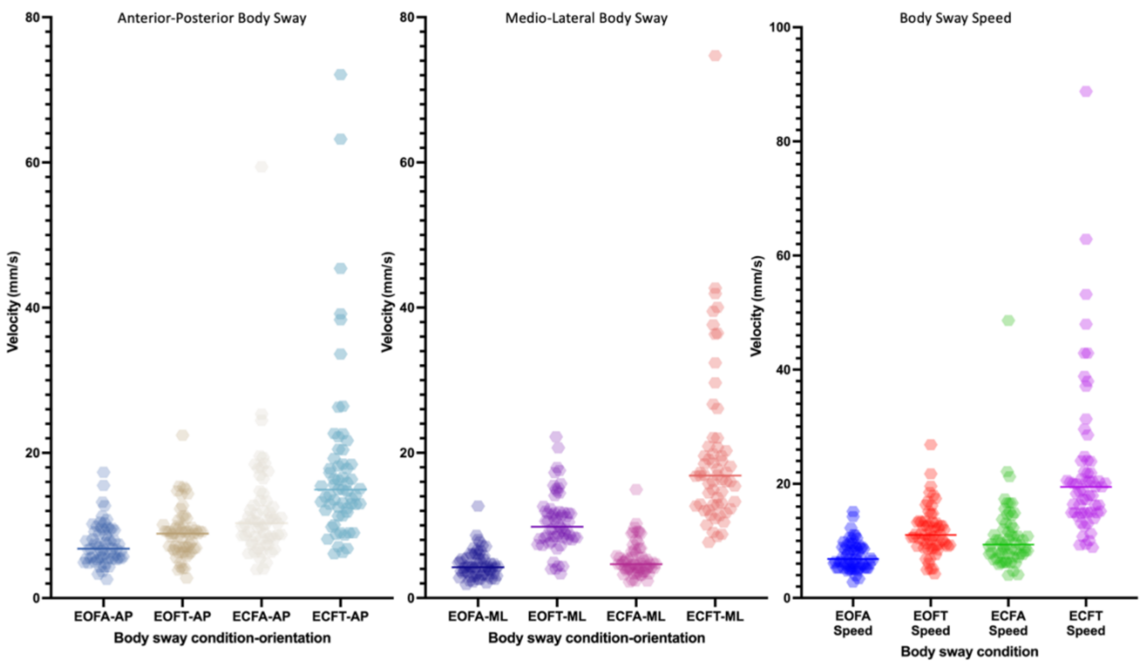
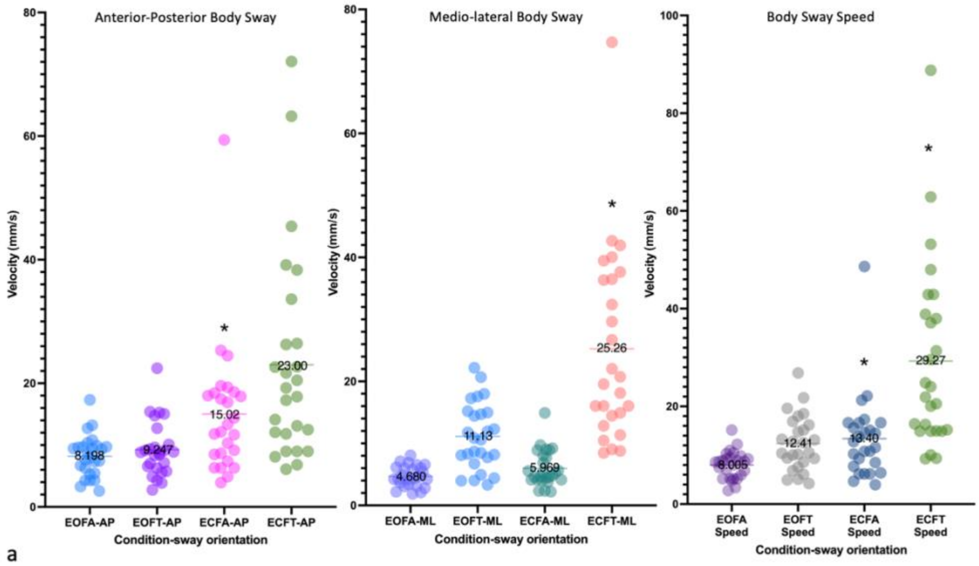


Figure 5.9: Body sway data of all participants. The conditions tested were Eyes open-feet apart (EOFA), eyes open-feet together (EOFT), eyes closed-feet apart (ECFA), and eyes closed-feet together (ECFT), in random order (bars indicate median values). For all four conditions, three outcome measures were analysed: (from the left) anterior-posterior "AP" and medio-lateral "ML" centre of pressure velocity, in addition to centre of pressure "Speed". As can been seen in the three graphs, data appears to be positively skewed, and so median was more informative of the average BSw value of participants than the mean. For all three outcome measures (AP, ML and Speed) it was found that BSw scores were at their lowest for the EOFA condition, and highest for the ECFT condition. Upon analysis, the difference between conditions was found to be significant for all three outcome measures: AP body sway (p=0.001), ML body sway (p=0.001) and speed (p=0.001).

When examining the data in terms of BSw direction, it appears that on average (Table 5.1), body sway MLV was lowest during feet apart (EOFA, ECFA) compared to feet together (EOFT, ECFT) whether eyes were open or closed. Body sway APV on the other hand tended to be lower when eyes were open (EOFA, EOFT) compared to when eyes were closed (ECFA, ECFT). An initial Friedman test confirmed that there was a significant difference between direction of body sway (APV and MLV) for all conditions ($\chi^2_{(7)}$ = 367.2, *p<0.001*). We then went on and analysed the data between APV and MLV body sway conditions separately using a Wilcoxon Signed-Ranks test and confirmed that there was an effect of body sway direction between each of the four conditions (*all p<0.001*): Specifically, MLV sway was significantly lower than APV sway during feet apart, regardless of vision condition (z=-6.62, *p<0.001*; z=-6.78, *p<0.001* respectively for EOFA and ECFA); and APV sway was significantly lower during feet together, also for both vision conditions (z=-4.74, *p<0.001*; z=-3.69, *p<0.001* respectively for EOFT and ECFT).

We then went on to explore the effect of condition on BSw scores for each of the BSw outcome measures (APV, MLV and speed of body sway): A Friedman test confirmed there was a significant effect of condition for all three outcome measures: APV ($\chi^2_{(3)}$=145.3, *p<0.001*), MLV velocity ($\chi^2_{(3)}$=161.8, *p<0.001*) and speed ($\chi^2_{(3)}$=160.8, *p<0.001*).

Body sway data was then divided and analysed based on age group (Figure 5.10a-5.10b):



a

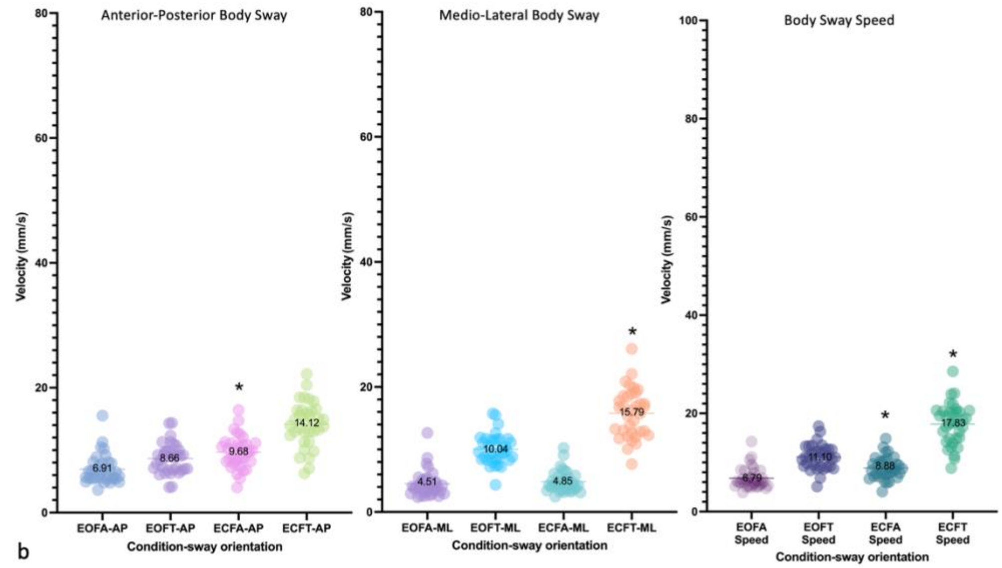


Figure 5.10: Body sway data for (a) old and (b) young age groups. The conditions tested were Eyes open-feet apart (EOFA), eyes open-feet together (EOFT), eyes closed-feet apart (ECFA), and eyes closed-feet together (ECFT), in random order (bars indicate mean values). For all four conditions, three outcome measures were analysed: (from the left, both 5.10a and 5.10b) anterior-posterior "AP" and medio-lateral "ML" centre of pressure velocity, in addition to centre of pressure "Speed". At first glance, all three outcome measures (AP, ML, and speed) appear to show lower values for the young age group compared to the older age group. However, upon analysis it was found that the difference in body sway between age groups was only significant (*) for ECFA-AP (p=0.028), ECFT-ML (p=0.015), ECFA Speed (p=0.014), and ECFT Speed (p=0.01).

* Difference in body sway outcome measure is significant between age groups at the 0.05 level (two-tailed).

We initially performed a general multivariate test to look for any interactions between variables within our data. We found a significant interaction between age and vision during APV (*p=0.043*) MLV (*p<0.001*) and speed (*p=0.002*) of body sway during the eyes closed conditions only.

For further clarification, a Kruskal-Wallis test was performed between old and young participants for each of the four testing conditions. Results indicated that there was a significant effect of age on BSw scores for

ECFA-AP (H (1)=5.45, *p =0.02*), ECFT-ML (H (1)=4.53, *p=0.033*), ECFA-speed (H (1)=6.51, *p=0.011*), and ECFT-speed (H (1)= 5.24, *p=0.022*). This indicates that vision always affected speed of BSw between age groups, regardless of foot placement. Looking at the graph (Figure 5.10) this seems to imply that older participants were more likely to sway faster than younger participants when their eyes were closed, meaning they were more likely than the younger participants to use vision to maintain their balance.

c. Reaction time and body sway:

After examining RTm and BSw data separately, we looked for any potential correlations between the two, to see whether RTm could predict balance performance measured via BSw. As the only shared velocity between the two age groups was 0.75°/s, we analysed this outcome measure against our BSw outcome measures: Results indicated a moderate relationship between RTm and AP BSw, as well as RTm and BSw speed:

A spearman correlation indicated RTm was positively correlated with EOFA-AP (r(60)=0.352, 95% CI [0.1-0.56], *p=0.005*), ECFA-AP (r(60)=0.36, 95% CI [0.11-0.57], *p=0.004*), EOFA-speed (r(60)=0.351, 95% CI [0.1-0.56], *p=0.006*), and ECFA-speed (r(60)=0.356, 95% CI [0.11-0.56], *p=0.005*).

Figure 5.11 illustrates the relationship between RTm and AP-BSw during eyes open (5.11a) and eyes closed (5.11b), while Figure 5.12 illustrates the relationship between RTm and BSw speed during eyes open (5.12a) and eye closed (5.12b):

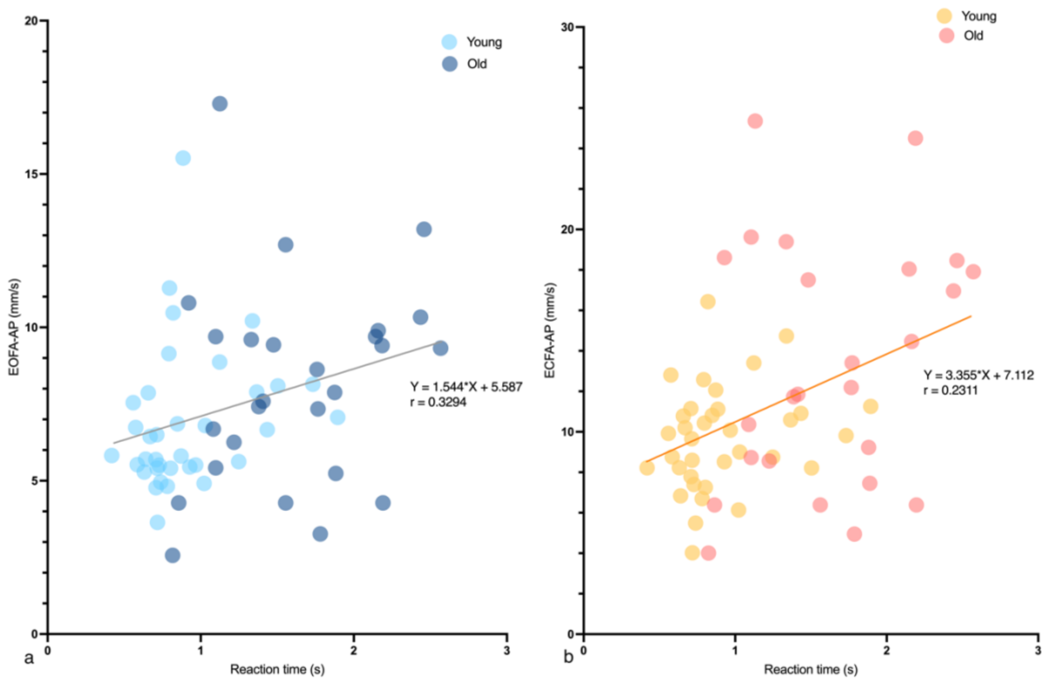


Figure 5.11: Correlation between reaction time and anterior-posterior (AP) body sway. Body sway conditions shown are (a) eyes open-feet apart (EOFA) and (b) eyes closed-feet apart (ECFA). Individual scores are colour-coded for each graph to compare young and old aged participants. It is clear that young participants tended to score in the bottom left quartile in both graphs compared to the old participants indicating the former scored lower for both reaction time and body sway. A significant correlation was found between (a) reaction time and EOFA-AP (r(60)=0.352, 95% CI [0.1-0.56], p=0.005) as well as between (b) reaction time and ECFA-AP (r(60)=0.36, 95% CI [0.11-0.57], p=0.004).

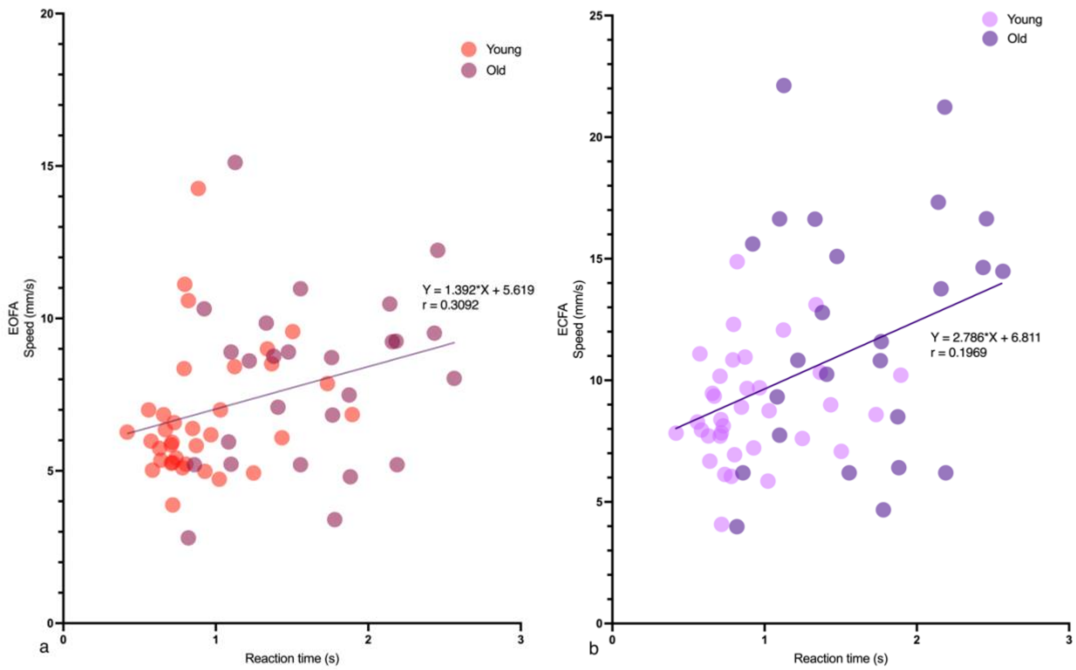


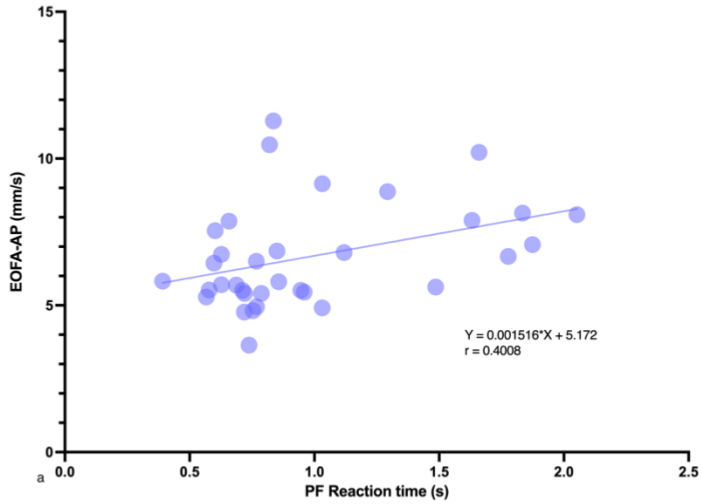
Figure 5.12: Correlation between reaction time and body sway speed. Body sway conditions shown are (a) eyes open-feet apart (EOFA) and (b) eyes closed-feet apart (ECFA). Individual scores are colour-coded for each graph to compare young and old aged participants. For the eyes open condition (a), young participants tended to score towards the left side of the graph, signifying a lower reaction time, but not a necessarily lower body sway than their older counterparts. For the eyes closed condition (b), we see a similar trend as was seen above in Figures 11a-b, with young participants scoring in the lower left quartile of the graph, indicating lower scores for both reaction time and body sway. A significant correlation was found between both (a) reaction time and EOFA-Speed (r(60)=0.351, 95% CI [0.1-0.56], p=0.006) and (b) reaction time and ECFA-Speed (r(60)=0.356, 95% CI [0.11-0.56], p=0.005).

To somewhat clarify the effect of age on our outcome measures, we analysed the relationship between age and RTm, as well as age and BSw, restricting our analyses to the BSw conditions in which we found a significant effect with RTm (Figures 5.11 and 5.12). Using a Spearman correlation test, we found a moderate positive correlation between age and RTm (r(60)=0.63, 95% CI [0.44-0.76], *p<0.001*), however, no significant correlations were found between age and any of the aforementioned BSw outcome measures (*p>0.05*).

Relationship between body sway and reaction time based on direction of ankle rotation:

Dorsiflexion and plantarflexion RTm data for the young age group was then separately analysed against their corresponding BSw outcome measures. This was done to assess whether direction of ankle rotation (upwards or downwards) had any influence on the relationship between RTm and BSw.

A Spearman correlation test indicated a significant correlation between 0.75°/s RTm during plantarflexion (PF RTm) and two of the BSw outcome measures: with AP body sway ($r(33)=0.401$, 95% CI [0.062-0.66], *$p=0.019$*) and with body sway speed ($r(33)=0.344$, 95% CI [-0.004-0.617], *$p=0.047$*), both during eyes open-feet apart (EOFA-AP and EOFA-Speed respectively). Figure 5.13 illustrates the relationship between PF RTm scores and EOFA-AP BSw (5.13a) and between PF RTm scores and EOFA-Speed BSw:



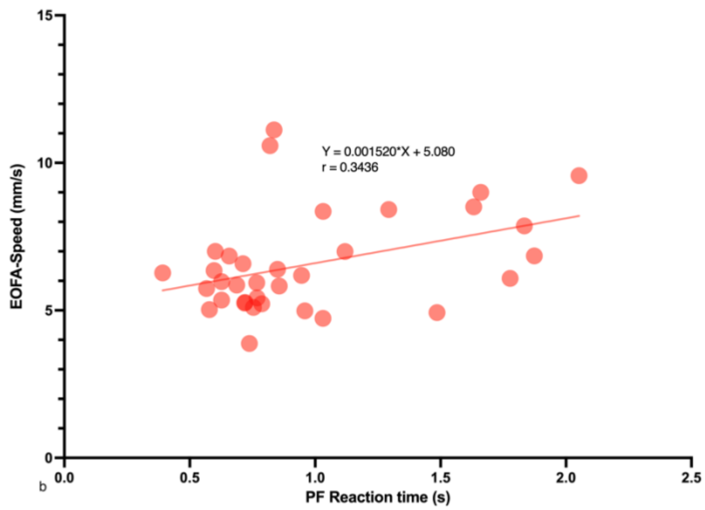


Figure 5.13: The relationship between plantar flexion (PF) reaction time and (a) anterior posterior body sway, and (b) body sway speed. Body sway condition for both graphs was eyes open feet-apart (EOFA) condition. Both body sway outcome measures (EOFA-AP and EOFA-Speed) each had a significant correlation with PF reaction time scores ($p<0.05$). Also depicted in both graphs is the line of best fit with its linear equation. These results may indicate that participants performance during downward rotation of the ankle was a better predictor of body sway performance than during upward rotation of the ankle.

5.5 Discussion

Two experiments were conducted using the proprioceptive RTm method we developed: the first was a preliminary study on 10 young participants, in which we tested how quick they were able to detect passive motion of their right ankle joint using 3 different angular velocities (0.25-0.5 and 0.75°/s). Results confirmed that velocity had a significant effect on RTm at the ankle, verifying that the higher the velocity, the smaller the RTm value, and vice versa. We then conducted our second study, to determine whether RTm at the ankle joint correlated with body sway during quiet standing. For this second experiment, we tested a comparably larger sample (n=61), divided into two age groups: young and old.

5.5.1 Reaction time and differences between age groups

Both age groups showed a similar trend with regard to RTm: When angular velocity of the footplate was increased, reaction time scores always decreased. Inferential testing confirmed a significant correlation for both age groups as well. When comparing data between young and old participants, young participants tended to score lower RTm scores compared to their older counterparts, and this difference was found to be significant. In addition, a spearman correlation test indicated a significant positive correlation between age and RTm indicating that RTm scores became higher (worse) when age increased. This is in line with our results from Chapter 3, in which a significant positive correlation between age and mean absolute error indicated that older aged participants tended to perform worse than younger aged participants when faced with an ankle joint position matching task. Similarly, a study by Yang et al. (2019) on ankle proprioception using the active standing extent discrimination apparatus (AMEDA) method during standing found that performance of the oldest age group in the study (75-90 years) was significantly lower than the other 5 age groups. In our study, ages of our older participants ranged from 69-86 years, indicating that not only were our results comparable to those of Yang et al. (2019), but also that our RTm method may be comparable to other known proprioceptive methods of testing in detecting an effect of age.

5.5.2 Differences in RTm scores based on ankle rotation direction

Another interesting finding from our RTm data was that the direction of ankle rotation seemed to affect participants performance. For both our preliminary and experimental studies, participants' RTm scores tended to be lower (better) during upward rotation, otherwise known as dorsiflexion. Although the difference in scores between dorsiflexion and plantarflexion was only significant in our experimental study, this may just be a matter of a higher number of participants (n=61 versus n=10) leading to greater statistical power. Moreover, a number of factors may have contributed to these findings: although the main goal with our experimental setup was to assess pure proprioception without the interference of other sensory input, such as tactile or vibratory sensation, it may be inevitable that they somehow affected results. For instance, the use of a footplate to passively move the foot during testing may have provided tactile cues in addition to proprioceptive input as to the direction of ankle rotation, specifically towards dorsiflexion. In addition, despite our efforts to dampen the sensation of vibration produced by the motor during rotation of the footplates, there may have been vibratory cues as to when the footplates started moving, and may have been more pronounced during upwards rotation due to the placement of the foot on the footplate. Conversely, it's plausible that results were indeed true, as calf muscle spindles are thought to be the main contributors to ankle proprioception during standing (Reynolds et al., 2020), and so any slight rotation towards dorsiflexion, which would cause a change in calf muscle length, may be more easily detectable than a rotation towards the opposite direction.

5.5.3 Body sway and differences between age groups

We assessed BSw to measure participants balance during quiet standing, during four different conditions: EOFA,EOFT, ECFA, ECFT. Our outcome measures were anterior posterior (AP), mediolateral (ML) velocity, and sway speed (mm/s). As expected, participants swayed the slowest during what was considered the easiest standing condition (EOFA) and fastest during what was considered the hardest of the four conditions (ECFT) for all three of our outcome measures. Analyses confirmed significant differences between each and every one of our conditions per outcome measure, indicating that our participants ability to balance during quiet standing became significantly more challenging with the absence of vision and the narrowing of stance.

While studying our BSw data in more detail, we noticed that there was an effect of stance width on body sway: When feet were apart, participants swayed significantly faster in the AP direction, whereas when feet were together, participants swayed significantly faster in the ML direction, regardless of vision (eyes open or closed). This finding is somewhat in line with those of Gatev et al. (1999), who determined that a narrow stance led to increased lateral movement of the centre of gravity (COG), whereas it did not increase movement of COG in the anterior-posterior direction. Add to that an earlier study by Day et al. (1993), who found that stance width exerted a greater influence on lateral velocity than it did on AP velocity. This slightly differs from our findings, as we noticed the increase in body sway velocity towards either the AP or ML direction contingent upon stance width: During the feet apart conditions, we found ML BSw velocity was significantly lower than AP BSw velocity. Whereas during the feet together conditions, AP BSw velocity was significantly lower than ML BSw velocity. An explanation for our finding may be the transition between ankle and hip strategies during narrow stance, as there is typically a decreased role of ankle mechanisms and an increased role of hip mechanisms in the AP plane with feet together, whereas in the ML plane, both ankle and hip mechanisms increase (Gatev et al., 1999).

However, when analysing our data based on age group, we found a significant interaction between age and vision for AP and ML BSw velocity, as well as for speed of BSw during the eyes closed conditions only. These results indicate that older participants rely more on vision than younger participants to maintain their balance. Upon further analysis, we found an effect of age only when eyes were closed and feet were apart (ECFA) for AP BSw velocity. This may indicate that with the absence of vision, elderly participants were less able to effectively rely on ankle, knee and hip strategies to maintain standing balance than young participants when feet were apart. However, when feet were together, we found no significant difference between age groups' AP BSw. Looking at ML BSw velocity, we also found an effect of age, however this time when eyes were closed and feet were together (ECFT). This may also be attributed to difficulties in regulating balance mechanisms between the ankle and hip, which, as mentioned above, typically increase during narrow stance. When we analysed BSw speed, we found an effect of age whenever vision was absent: this included both conditions of stance width (ECFA) and (ECFT). This makes sense, as speed here combines AP and ML BSw velocities to produce a directionless measure of velocity of COP. As such, speed as an outcome measure may not be as informative as AP and ML velocity, but may be used here for a more general interpretation. Nonetheless, our results indicate a significant age-related increase in body sway with eyes closed, which seems to imply that older participants tend to rely more on visual information than younger participants, possibly due to declines in peripheral sensory perception associated with aging (Jeka et al., 2010; Franz et al., 2015). Numerous studies support our findings, underscoring the tendency of older adults to rely more on visual input than other sensory systems to control upright standing (Wade et al., 1995; Borger et al., 1999; Simoneau et al., 1999; Yeh et al., 2014). Sensory reweighting is the process we believe may explain this observation, as it involves the brain preferring more reliable sensory cues, in this case vision, to determine the direction of vertical when other sensory cues may be deficit or not as reliable as they once were due to ageing or disease (Alberts et al., 2019). As such, it may be reasonable to say that age seems to affect proprioception and vestibular function, thereby creating greater reliance on vision.

5.5.4 Correlation of RTm and BSw

The main findings of our experiment were the moderate correlations we found between between 0.75°/s RTm and BSw. However, the correlations we found were specific to the BSw conditions in which feet were apart for APV and speed (Figures 5.11 and 5.12). As the positioning of participants feet during the RTm experiment was very similar to that used during the BSw feet apart conditions (standing position with distance between big toes 20 cm), in addition to a similar plane of ankle movement during both experiments (AP/sagittal plane) it makes sense to find a correlation between the two measures. Our results indicate that ankle proprioceptive reaction time is significantly correlated with AP body sway with feet apart, for both eyes open and eyes closed conditions. Although we found an effect of age on RTm scores, with older participants scoring significantly larger reactions times than younger participants, the same cannot be said of BSw, as no correlations were found between age and any of our BSw conditions. This suggests that age did not dictate how well participants were able to balance during our quiet standing conditions, which may indicate that declined ankle proprioception is associated with worse postural balance in adults of all ages. Several studies observed similar findings: A robust study by Deshpande et al. (2016) found that ankle proprioceptive acuity, measured by TTDPM, had a significant graded relationship with objective and self-report measures of balance and mobility, meaning that a decline in proprioceptive performance was associated with poor balance in adults of all ages. Another study by Chen and Qu (2019), found that JPR error had a significant correlation with AP and ML COP body sway scores in both young and older adults. In addition, Wang and Fu (2022) found that ankle proprioceptive threshold was correlated with balance control in older aged adults (60-80 years), but not in the oldest age group (>80 years). Lastly, a study by (Song et al., 2021) also found a correlation between ankle proprioceptive threshold and balance when balance was assessed using the Berg Balance Scale (BBS) but not when it was assessed using a force plate. A noticeable limitation of all four mentioned studies was that proprioception of the ankle was measured in sitting, which according to Refshauge and Fitzpatrick (1995) is not the ideal position to test ankle proprioception. In addition, their results cannot be directly compared with ours, as the methods used to measure proprioception and balance were numerous, a notorious issue with proprioception studies. To that effect, a limitation of our study was the limited amount of literature on proprioceptive RTm with which we can compare and contrast our findings. However, for this same reason, we believe our results somewhat add to the gap in proprioception studies and hopefully may provide some insight for future experiments for ourselves and others.

5.5.5 Plantarflexion RTm and BSw

An interesting finding was the moderate correlation between RTm during downward rotation and APV BSw in addition to BSw speed. No correlations were found between upward rotation and any of the BSw conditions. Both Song et al. (2021) and Wang and Fu (2022) demonstrated similar findings, with correlations between ankle proprioception and standing balance limited to the plantarflexion direction. They found no correlations between proprioception during ankle dorsiflexion and balance measures. A possible explanation for this may be that individuals rely more on input from the plantar flexor muscles, or calf muscles, during balance tasks (Wang and Fu, 2022) as calf muscle spindles are arguably the main contributors of ankle proprioception information during standing (Reynolds et al., 2020). Clinically, this may imply that proprioceptive acuity during plantarflexion may be more predictive of a patient's balance ability in standing.

5.5.6 Limitations

In addition to the deficit of studies on proprioceptive reaction time with which to compare our methods and findings with mentioned above, the age population of our participants lacked middle aged adults (40-60 years). Inclusion of this age group may have strengthened some of our findings, or provided additional information to our study.

5.6 Conclusion

Our results indicate that ankle proprioceptive reaction time is significantly correlated with AP body sway with feet apart, for both eyes open and eyes closed conditions. We are therefore able to accept our alternative hypothesis "Proprioceptive acuity at the ankle as measured by reaction time threshold shows a significant relationship with quiet standing balance as measured by bipedal body sway". Our findings have direct clinical implications, as they add a bit more insight to the ever-growing body of evidence on the relationship between ankle proprioception and standing balance and may have some impact on treatment protocols catered to those with balance impairments. Our study was limited to quiet standing or static balance, so a natural progression for future studies would be experiments including dynamic balance tasks. Including middle aged adults for future studies may also add to the robustness of any findings of future experiments.

Chapter 6: General Discussion

Abstract:

For our general discussion, we present the main findings from our experimental chapters after which we compare and contrast them to each other and to related studies from recent literature to interpret how these findings may contribute to future experiments on ankle proprioception. Among the topics we discuss are the effect of ageing on proprioception and balance, the influence of muscle strength and muscle contraction on proprioception, and the relationship between balance and proprioception. We conclude our chapter with a section on future research opportunities and clinical implications related to our findings.

6.1 Introduction

Our research question "Which proprioception assessment method best predicts balance ability in healthy young and elderly populations?"—like proprioception itself—is a complex, multifaceted concept that we now believe cannot be answered through a single thesis. However, our research did seem to partly uncover some valuable information concerning ankle proprioception and its relationship with standing balance.

6.2 Main findings of our experimental studies-Recap

A large portion of our experimental process was used to establish suitable and sound methods to test ankle proprioceptive acuity to be able to explore the relationship between ankle proprioception and quiet standing balance. An appropriate method to test ankle joint position reproduction (JPR) was already in place, so we went on to develop methods to test proprioceptive amplitude threshold (°) and reaction-time threshold (s) at the ankle. The experiments we performed using the aforementioned methods are presented in Chapters 3, 4, and 5 respectively. Our first experiment (Chapter 3: The Effect of Age and Strength upon Joint Position Reproduction) explored the relationship between contractile and sensory muscular processes during an ankle joint position matching task. We also sought to find out whether a compensatory mechanism between these processes came into play for older-aged participants. Although our findings did seem to indicate an influence of muscle strength on proprioceptive ability for the two older aged groups (middle-aged and old), results only reached significance for the middle-aged group, where higher strength was associated with lower absolute error during the ankle joint matching task. However, it was confirmed that older participants performed significantly worse during JPR than younger participants. For our next experiment (Chapter 4: Active and Passive Detection of Threshold) we aimed to determine whether a relationship existed between active muscle contraction and proprioceptive acuity at the ankle during standing, measured using an amplitude threshold proprioceptive task. To do this, we analysed the effect of holding an isometric contraction at the calf muscle while performing the proprioceptive threshold task and compared our data with those gathered while performing the same task with a passive muscle condition (relaxed calf muscle). Results indicated that active muscle contraction did not directly affect proprioceptive acuity. However, participants torque variability during testing showed a significant correlation with detection threshold scores. Our findings suggest that decreasing muscle activity variability, essentially any background noise within the calf muscle, allows for enhanced proprioceptive acuity at the ankle joint. For our final experiment (Chapter 5: Proprioceptive Reaction Time as a Predictor of Body Sway) we sought to determine whether proprioceptive acuity at the ankle correlates with body sway during quiet standing for young and elderly adults. We used a proprioceptive reaction time task to measure ankle proprioceptive acuity, and a standard force plate to measure quiet unperturbed standing balance during four conditions which ranged in difficulty. These conditions—from the easiest to the most difficult—were (1) eyes open, feet apart, (2) eyes open, feet-together, (3) eyes closed, feet-apart, and lastly (4) eyes closed, feet-together. Our results indicated that proprioceptive reaction time at the ankle was significantly correlated with anterior-posterior centre of pressure (COP) velocity during the feet apart conditions. A significant correlation was also found between reaction time during plantarflexion specifically and anterior-posterior COP velocity during eyes open, feet apart. We also found that age was significantly correlated with reaction time, with older individuals scoring worse than younger aged individuals. We also found an effect of age on body sway COP speed but only when vision was occluded. In the next sections, we will go on to discuss several issues based on our findings, including ageing and its effect on proprioception, the influence of muscle strength and muscle contraction on proprioception and the differences between them, and lastly, the relationship between proprioception and balance.

6.3 Ageing: How does it affect proprioception and balance?

A better understanding of the effect of age on ankle proprioception—and its implications on postural control— is very relevant at present, considering the growing percentage of the world population over the age of 65 (Ageing | United Nations, n.d.). Having said that, the mechanisms in which ageing effects proprioceptive acuity has been a topic of discussion over the last few decades, with the majority of literature indicating that ageing is almost always associated with deterioration of our proprioceptive sense (Ribeiro and Oliveira, 2011), among other vital senses we use in our daily lives, such as our visual and auditory senses. For our experiments, two proprioceptive testing methods were applied to older aged participants (≥70 years) to assess their ankle proprioceptive acuity: Joint position reproduction (JPR) and reaction time (RTm). Our findings from both experiments indicated that age was inversely related to ankle proprioception at the ankle. Several studies showed similar results between ageing and JPR at the ankle, such as Westlake et al. (2007) and Deshpande et al. (2003). However, the studies we found on ageing and proprioceptive RTm indicated there was no relationship found between the two (Verschueren et al., 2002; Djajadikarta et al., 2020). We had intended to assess older aged participants using our detection threshold method in addition to the two aforementioned methods, however the experiment proved to be too time consuming to be practical for more mature participants. Nonetheless, several studies using a detection threshold method (Verschueren et al., 2002; Deshpande et al., 2003; Westlake et al., 2007) concluded a significant relationship between detection threshold and ageing.

As for the study by Djajadikarta et al. (2020), they also assessed two other ankle proprioception testing methods alongside RTm: JPR and detection threshold. While their results indicated that age had no effect on ankle proprioception when movement history was controlled, several key details of the methods used in this study may be argued as not being satisfactory. For the amplitude detection threshold for example, the angular velocity used to passively move the ankle was 13°/s, which in our opinion may have produced a ceiling effect, as it seems far too rapid a movement for a detection threshold experiment. For our threshold experiment (Chapter 4) we used velocities a fraction of this number (eg. 0.1, 0.25 and 0.5°/s) and found that differences between participants' threshold scores become less apparent the faster the velocity, or in other words threshold scores became less variable the faster we moved the footplate. Although we only tested younger aged participants (19-38 years) and are unaware of how our results would have translated if older adults were recruited, given our present findings an angular velocity of 13°/s would unlikely be sensitive enough to capture differences in threshold between participants. The same may be said for the angular velocities used for the RTm method in both the Verschueren et al. (2002) and Djajadikarta et al. (2020) studies, which were 30°/s and again 13°/s respectively. In addition, for the JPR assessment used by Djajadikarta et al. (2020), angles chosen for testing (-10°, -5°, 0°, 5°, 10°) may have been spaced too wide apart to detect an effect of age, as our own findings from the JPR experiment in Chapter 3 indicated a significant effect of age on absolute error at angles 0° and 2° only, with no effect at the other larger angles tested. In contrast, a strength of the Djajadikarta et al. (2020) study was controlling for movement history between each and every test trial for all the three methods used, a detail we applied for our JPR and threshold detection studies but not for our RTm study. Controlling for movement history, given the effect of thixotropy on proprioceptive acuity (Proske and Gandevia, 2012) is an important detail that should be accounted for in most studies measuring proprioception, however it is our opinion that this was not the reason for the findings obtained by Djajadikarta et al. (2020). It is more likely that the velocities chosen for the 3 testing methods were inappropriate, the reason we believe the effect of age on proprioception was found to be negligible in this instance. Had the chosen velocities been more conservative, similar to those used by Refshauge and Fitzpatrick (1995)—from which we modelled our detection threshold method—it may have been more likely to observe an effect of age on ankle proprioception.

With regard to the effect of ageing on balance, our own results were not so clear cut. In our last experiment (Chapter 5), balance was measured via body sway centre of pressure (COP) velocity obtained during quiet standing on a force plate. Although the main objective of collecting this body sway data was to correlate it to our proprioceptive data, we also performed analyses between subjects to check for any trends regarding age. Our results indicated that older participants seem to rely on a compensatory mechanism to control quiet standing: The effect of age was only significant when vision was occluded, suggesting that they relied on vision significantly more than younger participants to maintain their balance during the four conditions we assessed. This suggests that proprioception acuity is somewhat decreased in elderly individuals, prompting the use of other more reliable sensory cues to maintain balance. Our findings echo those of numerous studies, in which the tendency of older adults to depend on visual cues more than other sensory cues to control upright standing is highlighted (Wade et al., 1995; Borger et al., 1999; Simoneau et al., 1999; Yeh et al., 2014).

An additional factor that may lead to greater reliance on visual information related to diminished proprioceptive acuity in the elderly may be muscle weakness. Lower-limb muscle weakness is a common feature for many older adults (Frontera et al., 1991; Vandervoort, 2002) with an average 30-40% decrease in muscle strength for adults from 30-80 years of age, and further significant declines occurring with advanced age (Doherty et al., 1993; Brooks and Faulkner, 1994; Lindle et al., 1997). In a study by Butler et al. (2008) individuals with lower limb weakness relied more on vision to detect and control body sway than those with comparatively strong lower limb muscles. As such, muscle weakness may contribute further to older aged adults' reliance on vision. This may be explained by the increase in muscle activation required to maintain upright standing in older aged individuals (Billot et al., 2010) which may reduce proprioceptive acuity (Wise et al., 1998; Proske et al., 2000). It should be mentioned that strategies incorporating visual information have been shown to induce delayed—as well as less accurate—fall avoidance responses, contrary to adaptive strategies based on proprioceptive and vestibular information (Vouriot et al., 2004). In essence, this last fact alone should serve as incentive to increase physical activity in older adults, specifically that which focuses on improving muscular strength.

6.4 The influence of muscle strength and muscle contraction on proprioception

Regular exercise has been shown to attenuate the decline in proprioceptive acuity in older aged adults (Tsang and Hui-Chan, 2003; Adamo et al., 2009; Ribeiro and Oliveira, 2010; Wang et al., 2016; Niespodziński et al., 2018). This may be explained by the association between regular physical activity and increased muscular strength observed across the lifespan of both male and female adults (Sandler et al., 1991; Rantanen et al., 1997). Lower body strength in particular has been found to be significantly correlated with physical activity (Leblanc et al., 2015) emphasising the importance of incorporating physical activity into the daily lives of adults of all ages, specifically older aged adults with an increased risk of falling. Findings from our experiment on the effect of strength on proprioception (Chapter 3) showed that both middle aged (33-68 years) and older aged (69-90 years) groups with higher strength performed better than their lower strength counterparts in a joint matching task. However this effect was only significant for the middle aged group.

This may be explained by the type of participants recruited for our older age group: Healthy older adults that agree to participate in studies such as ours tend to be more “fit” than the general population of the same age, so there may not have been a natural spectrum of strength levels among recruited participants in this instance. This in turn may have implications on results when analysed alongside younger age groups with average fitness levels compared to the general age-matched population.

Nevertheless, several studies corroborate the idea that stronger muscles are associated with better proprioceptive acuity: Guney et al. (2016) found that quadriceps strength was significantly related to joint position sense at the knee; similarly, Hurler et al. (1998) found that quadriceps weakness was correlated with decreased joint position acuity. Unsurprisingly—and likely consequentially—muscle strength has also been shown to be linked to improved balance: The aforementioned study by Butler et al. (2008) found that subjects with weaker dorsiflexors displayed greater postural sway than those with stronger dorsiflexors when they were forced to rely upon their proprioceptive senses by occluding vision. Additionally, a study by Lord et al. (1993) found that muscle strength was an accurate predictor of body sway performance on an unstable surface.

In contrast, most of the studies we found exploring the relationship between voluntary muscle contraction and proprioceptive acuity indicated that active muscle contraction negatively affected proprioceptive acuity (Niessen et al., 2009; Kwon et al., 2013; Peters et al., 2017a). Although the findings from our active passive threshold experiment (Chapter 4) were somewhat similar to that of the literature, indicating that proprioceptive performance tended to be worse when participants held an isometric contraction at their calf muscle compared to when their lower limbs was relaxed, our results did not reach significance. Nevertheless, we did uncover an interesting trend within our data: Participants that were able to maintain steadier isometric contractions performed significantly better at the proprioceptive task. We believe that a steadier muscle contraction is likely associated with increased muscle strength, and—on hindsight—there may have been an opportunity to gain much more insight on the link between muscle strength and proprioceptive acuity had we measured our participants muscle strength during this experiment. We found a significant relationship between muscle strength and JPR in middle-aged adults in our previous experiment, so it would be interesting to explore whether we find similar results between muscle strength and threshold detection, or reaction time in the future.

6.5 The relationship between balance and proprioception

It is generally acknowledged that proprioception signals from lower limb muscles contribute the main source of information for postural control, with those arising from the ankle region in particular being the most important in terms of standing balance (Goble et al., 2011; Fling et al., 2014; Han et al., 2015; Henry and Baudry, 2019). This is due to their enhanced sensitivity in detecting body sway during unperturbed upright standing which mainly results from muscle length variations brought about by rotations around the ankle joint. Consequently, standing balance is the functional ability most affected by proprioceptive impairment (Inglis et al., 1994; Lord and Ward, 1994; Lajoie et al., 1996; Horak et al., 2009; Butler et al., 2014; Fling et al., 2014; Deshpande et al., 2016; Huzmeli et al., 2018; Cho and Kim, 2021; Labanca et al., 2021) which only highlights the importance of dissecting the nature of this relationship and determining the type of proprioceptive assessment that best predicts balance ability.

In our last experiment (Chapter 5) in which we studied the relationship between balance and proprioception—with quiet standing balance measured via body sway COP velocity, and proprioception measured via reaction time—we found similar results to those we observed in the literature: Proprioceptive reaction time at the ankle joint was found to be moderately correlated with body sway COP velocity in the anterior-posterior direction with feet apart. Vision did not effect this correlation, as the results were significant during both eyes open and eyes closed conditions. We also found a moderate correlation between RTm during plantarflexion and anterior-posterior BSw, a result which mirrors other recent studies (Song et al., 2021; Wang and Fu, 2022). Our findings imply not only that proprioceptive RTm may be considered a good predictor of quiet standing balance in adults of all ages, but also that RTm during plantarflexion may be used to predict a person's balance performance.

6.6 The contribution of "non-sensory" functions of the proprioceptive system

In our introduction, we mentioned that the proprioceptive system may be organized into "sensory" and "non-sensory" functions. Using this analogy, only the sensory function of the proprioceptive system may be referred to as "proprioception", while the "non-sensory" function is referred to as automatic reflexes or postural corrections: These are functions conveyed using the same mechanoreceptors that are stimulated during conscious proprioceptive tasks, but without direct awareness.

In the context of our experimentation method, the potential influence of non-sensory function of the proprioceptive system may be observed during our last experiment, when we measured participants' body sway during quiet standing. It is noteworthy to mention that our instructions to participants before commencing this test were to "relax and maintain quiet standing" whereas for all of our proprioceptive experiments, we instructed participants to report when they perceived (1) bilateral ankle joints to be in the same position (angle) during JPR trials (Chapter 3), and when they perceived (2) ankle motion had taken place during threshold detection and reaction time trials (Chapters 4 and 5 respectively). As such, we assume that any postural corrections that occurred during the body sway test were automatic, not requiring participants direct attention or awareness as to how their body was maintaining a quiet standing position throughout the trials.

This further solidifies the idea above that the proprioceptive system operates on two levels, one in which we are "aware of the mechanical and spatial condition of the body" and the other in which unconscious, automatic reflexes are responsible for maintaining upright balance.

6.7 Future studies and clinical implications

A single proprioceptive assessment method is unlikely to capture a person's overall proprioceptive ability. This is due to the complexity of proprioception, as different peripheral and central neural processes contribute to different aspects of our proprioceptive sensation; for example, proprioceptive reaction time (movement detection) requires different peripheral processes than say passive joint position matching, in which the target limb may remain in place while the contralateral limb is passively rotated into position, which also differs from the central and peripheral processes required if the joint matching is based on memory (target limb does not remain at target).

Having said that, it was our intention to investigate the link between proprioception and function by comparing and contrasting the relationship between several types of ankle proprioceptive assessments with body sway data, to determine which test best predicted standing balance; however this was not possible due to time constraints and personal circumstances. Nevertheless, we invite fellow researchers to continue in our experimental path in hopes of further clarification on the matter. In addition to comparing other proprioceptive methods to balance, we also believe it important to explore the effect of muscle strength on proprioception in more depth, by researching which proprioceptive assessment method is most affected by muscle strength for example. Also, whether increased muscle strength has an effect on age-related proprioceptive deficits as measured by detection threshold or reaction time.

Findings such as these may allow rehabilitative clinical practices to be less time consuming for patients and rehabilitative therapists alike by focusing on assessments and exercises that are proven to be informative of a patients' physical status, and in our case, their ability to balance during quiet standing. Additional research is needed to effectively convert these findings into practical methods that are applicable in a clinical setting, as there is now convincing empirical evidence that approaches such as active movement and proprioceptive exercise can lead to large gains in both balance control and motor function (Beckerle et al., 2022).

6.8 Concluding remarks

Our research was focused on different methods of proprioceptive assessment at the ankle joint and how results were affected by age, muscle strength and the state of the muscle (relaxed or contracted) and how they related to standing balance. We found that age had a significant effect on proprioception as measured by JPR and RTm, and on body sway when vision was occluded. We also found a significant effect of muscle strength on proprioceptive performance, in which higher strength was associated with lower joint position errors for middle aged adults. In addition, we found that the ability to effectively hold an isometric contraction was associated with better proprioceptive acuity. More research is needed in this area to determine how muscle strength comes into play in this instance. Last but not least, we found that reaction time at the ankle joint effectively predicts anterior-posterior body sway for adults of all ages, and when analysed further, it was found that plantarflexion reaction time predicted anterior-posterior body sway better than dorsiflexion reaction time. However, additional research is needed to determine which proprioceptive assessment method best predicts balance ability.

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Appendices

Target angle	-6°	-4°	-2°	0°	2°	4°	6°
Young (20-32 yrs.)							
Mean (°)	2.776	2.201	1.758	1.266	1.231	1.959	2.163
Std. Deviation	1.175	1.144	0.8995	0.5788	0.3477	0.6959	0.8874
Middle-Aged (33-68 yrs.)							
Mean (°)	3.515	2.633	1.663	1.273	1.322	2.399	3.398
Std. Deviation	1.257	0.8355	0.5896	0.4761	0.5964	0.8205	0.9572
Old (69-90 yrs.)							
Mean (°)	3.788	2.761	2.365	1.898	1.885	2.468	2.989

Std. Deviation	1.509	1.448	1.603	0.9225	0.5204	0.9731	1.285

Appendix 1: A comparison between age groups.
Mean and standard deviation of absolute error recorded at each target angle.

Target angle	-6°	-4°	-2°	0°	2°	4°	6°
Constant Error (°)	3.025	2.170	1.400	0.2784	-0.6212	-1.825	-2.505

Appendix 2: Mean constant error (CE) at each target angle.
CE was lowest at 0° increasing proportionally with the increase in angle both ways.

Target angle	**-6°**	**-4°**	**-2°**	**0°**	**2°**	**4°**	**6°**
Young (20-32 yrs.)							
Mean (°)	-3.87	-2.33	-1	0.22	1.65	2.56	4.44
Std. Deviation	2.06	1.67	1.49	1.12	0.98	1.06	1.40
Middle-Aged (33-68 yrs.)							
Mean (°)	-2.46	-1.5	-0.68	0.12	0.99	1.77	2.76
Std. Deviation	1.40	1.19	1.01	1.11	0.87	0.96	1.02
Old (69-90 yrs.)							
Mean (°)	-2.37	-1.64	-0.21	0.44	1.48	2.26	3.56

Std. Deviation	1.61	1.50	1.69	1.64	1.51	1.67	1.84

Appendix 3: Mean and standard deviation of average left (matched) foot position for each target angle.
Young participants showed the highest accuracy in terms of mean foot position per angle while middle aged participants showed the least variation among target angles.

Target angle	**-6°**	**-4°**	**-2°**	**0°**	**2°**	**4°**	**6°**
Left Torque							
Mean ± Std. Deviation (N.m)	7.66 ± 5.16	8.48 ± 5.31	9.91 ± 5.92	11.05 ± 6.08	12.35 ± 6.69	13.43 ± 6.81	15.06 ± 7.32
Right Torque							
Mean ± Std. Deviation (N.m)	5.64 ± 4.54	6.77 ± 4.61	8.67 ± 4.87	10.67 ± 5.15	13.10 ± 5.74	15.21 ± 6.26	17.73 ± 6.93

Appendix 4: Mean and standard deviation values to compare between left and right torque.
Left torque is consistently higher than right torque on average.

Target angle		-6°	-4°	-2°	0°	2°	4°	6°
Young/High strength	Mean (°)	2.92	2.38	2.14	1.46	1.33	1.78	1.75
	St. Dev	1.431	1.522	1.157	0.7068	0.4382	0.8551	0.7332
Young/Low strength	Mean (°)	2.63	2.03	1.37	1.07	1.13	2.14	2.58
	St. Dev	0.9673	0.7002	0.2957	0.3859	0.2264	0.5038	0.8834
Middle/High strength	Mean (°)	2.56	2.08	1.38	1.40	1.53	2.51	3.29
	St. Dev	0.8079	0.4927	0.2789	0.5533	0.6677	0.9578	0.9280
Middle/Low strength	Mean (°)	4.47	3.18	1.95	1.15	1.11	2.29	3.51
	St. Dev	0.7821	0.7525	0.7043	0.3928	0.4819	0.7327	1.060

Old/High strength	Mean (°)	3.71	2.96	2.5	2.07	1.79	2.21	2.56
	St. Dev	1.843	1.706	1.283	0.6743	0.3601	1.097	1.181
Old/Low strength	Mean (°)	3.87	2.57	2.23	1.72	1.99	2.72	3.42
	St. Dev	1.265	1.270	1.991	1.160	0.6649	0.8496	1.343

Appendix 5: A comparison of leg strength on AE at each target angle for each age group.
Lower matching error scores are shaded in green and higher matching error scores in yellow for each age group separately.

Age groups	Mean Left Torque (all target angles) (N.m)
Young/High strength	9.86 ± 4.12
Young/Low strength	10.27 ± 5.72
Middle/High strength	9.69 ± 10.1
Middle/Low strength	11.67 ± 4.15
Old/High strength	10.4 ± 3.79
Old/Low strength	14.92 ± 7.18

Appendix 6: Mean left torque values for high and low strength groups among the three age groups.
The Lower torque value among each age group is highlighted.