Introduction

Falls are the cause of around 60% of all non-intentional injuries and 40% of accidental deaths in people aged 65 years and over [1]. As the majority of falls occur during locomotion [2] it is important to identify age-related gait changes that may predispose the elderly to falling. Trips are a major cause of falls during walking [2] and occur when the swing foot contacts an object or the ground [3]. During level walking, a trip is most likely to occur at the time of minimum toe clearance ($time_{MTC}$), at which time the typical distance between the toe and the ground is approximately 15 mm [4-7]. Mean or median values of minimum toe clearance (MTC) are around 15 mm and not different between the healthy young and elderly [4, 6, 7]. Each day however, most people take thousands of steps without tripping or falling [8], and therefore a measure of MTC central tendency may not be the only appropriate measure for investigations aiming to gain insight into tripping during level walking. MTC exhibits stride-to-stride variability [5, 7, 9] which, given the small margin for error, must be minimised to avoid the occurrence of toe-ground contact events over the course of each day. The elderly have been reported to exhibit greater MTC variability than the young [7], although the biomechanical factors underlying this age-related increase in MTC variability are unknown.

The locomotor system can be modelled as a seven-link rigid body kinematic chain consisting of the pelvis and stance and swing thigh, shank and foot segments [9]. The position of the swing toe is a direct function of the global position of the stance toe, the global orientation of the stance foot, the segmental lengths, and the 3D joint angles of the stance and swing leg hip, knee and ankle joints [9]. Many previous studies have reported age-related differences in gait kinematics exist across the gait cycle [4, 6, 10, 11] and while it is reasonable to suggest that such differences would be present at $time_{MTC}$, this has not been established. Furthermore, it could be expected that stride-to-stride fluctuations of the degrees-of-freedom (DOFs) within the lower body kinematic chain would be positively related with MTC variability. Winter [9] employed a sensitivity analysis approach to assess the effect of systematic manipulation of stance and swing leg joint angles on MTC and indentified that MTC was highly sensitive to some kinematic DOFs, e.g., stance hip adduction-abduction [9], but relatively insensitive to fluctuations in others, e.g., stance knee flexion-extension. Despite these theoretical findings, little is known about the empirical variability exhibited by the stance and swing leg kinematic DOFs at $time_{MTC}$, nor whether variability of individual kinematic DOFs is related to MTC variability.

The variability of a number of biomechanical gait variables have been reported to be greater in the elderly compared with the young [7, 12, 13]. The greater levels of MTC variability of the elderly compared with the young [7] may result from an age-related increase in variability of the kinematic DOFs within the kinematic chain at $time_{MTC}$, however this has not been established. A thorough investigation into the effect of ageing on bilateral joint kinematics and kinematic variability at $time_{MTC}$ may provide insight into the biomechanical factors responsible for trip-related falls in the elderly.

The purpose of this study was to examine the central tendency and dispersion of MTC and bilateral lower body kinematics at $time_{MTC}$ in young and elderly adults during level walking. It was hypothesised that, i) the young and elderly would exhibit differences in lower body kinematic DOFs at $time_{MTC}$, ii) MTC variability and lower body kinematic variability at $time_{MTC}$ would be greater in the elderly compared with the young, and iii) lower body kinematic variability would be positively correlated with MTC variability.

Methodology

Participant details

Ten young (age = 25.8 ± 3.1 yrs, height = 1.76 ± 0.07 m & body mass = 74.4 ± 9.1 kg) and nine elderly (age = 71.1 ± 3.4 yrs, height = 1.72 ± 0.06 m & body mass = 82.7 ± 11.6 kg) men participated in the study. All participants, i) were currently living in the community, ii) had normal or corrected to normal vision, iii) were free from any diagnosed musculoskeletal or neurological abnormalities, iv) required no mechanical aids or devices for walking, v) were able to walk continuously for 30-minutes, and, vi) had not experienced a fall-related injury in the previous two years. These exclusion criteria were applied to avoid potential confounding influences on gait patterns that were not reflective of ageing per se. The experimental protocol was approved by the Griffith University Human Research Ethics Committee, and all individuals provided written informed consent prior to participation.

Data collection and processing

Walking trials were performed on a custom built wooden framed treadmill with a walking surface 4.2 m long length and 1.6 m wide (Payne Engineering, Sydney, Australia). In order to avoid the steel reinforcement in the laboratory floor distorting electromagnetic tracking system (ETS) measurements, the treadmill was raised such that the walking surface was 1.32 m above the laboratory floor. Firstly, participants underwent a treadmill walking familiarisation period, which involved self-selection of their natural walking speed and walking at that speed for a 10-minute period. ETS sensors were then attached to the participants' pelvis, thighs, shanks and feet and custom heel and toe footswitches [14] were inserted into their footwear. Sensors were positioned to maximise coupling with underlying skeletal features. The participants then underwent a series of calibration trials, as described previously [15], and a 20-minute self-selected walking speed trial. Position and orientation data were sampled at 30.07 Hz using a FasTrak (Polhemus, Vermont, USA) ETS in conjunction with 6D-Research software (Advanced Motion Measurement, Arizona, USA) and resampled to 120 Hz using the technique described by Hamill et al. [16] to improve temporal and spatial resolution of gait data. Data were filtered using a zero-lag Butterworth second order low pass filter with a cut-off frequency of 6 Hz. 3D segment and joint angles were modelled using our previously described procedure [15], with the addition of a dynamic femoral reference frame optimisation procedure [17] to minimise cross-talk in knee joint kinematics. From the static trial, a virtual swing toe was defined within each foot technical coordinate system (TCS) as the point on the sole of the shoe directly inferior to the midpoint of the medial aspect of the first and the lateral aspect of the fifth metatarsal heads. The virtual swing toes were transformed from its foot TCSs to the global coordinate system for the gait trial. Footswitch signals were sampled at 500 Hz using an MP100 amplifier and associated AcqKnowledge software (Biopac Systems Inc, California, USA). Foot contact and foot off events were defined from the footswitch signals using a previously described algorithm [14].

One thousand consecutive left and right strides were analysed for each participant. MTC was defined as the first sample within each swing phase at which a minimum vertical toe position occurred. $Time_{\mathrm{MTC}}$ was reported as a percentage of the swing phase. Spatiotemporal variables, stance and swing leg joint angles, global pelvis angles and MTC were identified from each swing phase. With the exception of kinematic modelling, which was performed using custom Bodybuilder routines (Oxford Metrics, Oxford, UK), data were processed using routines developed in Matlab 7.0 (Mathworks, Natick, USA).

Statistical Analysis

Prior to hypothesis testing, the Shapiro-Wilk test was used to assess whether within-subject kinematics fitted a normal distribution. Of the 912 distributions assessed, only 227 met the criteria for normality. Thus, within-subject medians and inter-quartile ranges (IQR), i.e., the range between the first and third quartiles, were used as measures of central tendency and variability, respectively, as recommended for non-normally distributed data

[18]. Spatiotemporal variables and MTC median and IQR values was assessed using a general linear model (GLM) analysis with age group as a between-subject factor. The effect of age group on stance foot angle medians and IQRs, and stance and swing leg joint angle medians were assessed using separate multivariate GLM analyses, each with age group (young vs. elderly) as a between-subject factor, and stance foot, stance leg joint angles and swing leg joint angles, as the composite dependent variable. Stance and swing leg joint angle IQRs were analysed using a multivariate GLM with age group (young vs. elderly) as a between-subject factor and joint angle IQRs as the composite dependent variables. A multivariate approach was utilised to minimise the risk of Type II error [19]. Where a significant main effect of age group or an interaction involving age group was identified from multivariate analysis, univariate GLMs were performed to identify which segment/joint angles were different between the age groups. One-tailed Pearson's product moment correlations were used to assess the relationship between MTC variability and stance foot and joint angle variability at $time_{\mathrm{MTC}}$. Statistical analyses were performed using SAS Version 8 (SAS Institute Inc., Cary, NC, USA) with α level of $P<0.05$. Cohen's effect sizes were calculated for all age group comparisons [20]. Dependent variables are reported as the age group mean ± standard deviation.

Results

Subject characteristics and spatiotemporal variables

No age-related differences in height or body mass were identified. Spatiotemporal gait variables for the young and elderly participants are summarised in Table 1. Small but statistically significant temporal differences were identified between the age groups, with the elderly exhibiting a decreased stride duration, $F_{1,17}=4.61$; $P=0.047$, decreased percentage of the cycle in the stance phase, $F_{1,17}=6.55$; $P=0.020$, and increased percentage of the cycle in the swing phase, $F_{1,17}=6.46$; $P=0.021$, compared with the young.

[Insert Table 1 about here]

Median kinematics

Vertical swing toe trajectories for representative young and elderly subjects over 100 consecutive swing phases are presented in Figure 1a and c, respectively. As outlined in Table 1, there were no age-related differences in median MTC or median stance foot orientation at $time_{\mathrm{MTC}}$. The elderly displayed less stance hip extension, $F_{1,17}=11.88$; $P=0.003$, greater swing hip flexion, $F_{1,17}=7.90$; $P=0.012$, and less stance hip adduction $F_{1,17}=6.20$;

P=0.023, than the young. Stance and swing leg joint angles for the young and elderly subjects at $time_{MTC}$ are presented in Table 2.

[Insert Figure 1 & Table 2 about here]

Kinematic variability

Representative MTC values for representative young and elderly participants are presented in Figure 1b and d, respectively. MTC variability was greater for the elderly than the young, $F_{1,17}$=4.45; P=0.049, Table 1.

No age-related differences in within-subject variability of stance foot angles or joint angles were identified, however stance hip flexion-extension and adduction-abduction, stance knee adduction-abduction and internal-external rotation and swing knee adduction-abduction had moderate to large effect sizes suggesting greater variability for the elderly.

[Insert Table 3 about here]

Relationships between MTC variability and angular variability

MTC variability had a positive linear relationship with variability of a number of joint angles for the young and elderly groups, Figure 2. For the young group, MTC variability was positively correlated with knee adduction-abduction, r=0.661; P=0.019, ankle plantar-dorsiflexion, r=0.695; P=0.013, and ankle adduction-abduction, r=0.617; P=0.029 of the stance leg and knee flexion-extension, r=0.690; P=0.014, ankle plantar-dorsiflexion, r=0.938; P<0.001. For the elderly group, MTC variability was positively correlated with ankle adduction-abduction of the stance leg, r=0.826; P=0.003, and knee flexion-extension, r=0.588; P=0.048, ankle plantar-dorsiflexion, r=0.645; P=0.031, and ankle internal-external rotation, r=0.593; P=0.047. No significant correlations between stance foot angle variability and MTC variability were identified.

[Insert Figure 2 about here]

Discussion

The primary aim of the present study was to examine the central tendency and dispersion of MTC and bilateral lower body kinematics at $time_{MTC}$ in young and elderly adults during level walking. Overall, the results demonstrated no differences in preferred speed gait velocity or stride length between the age groups. Stride duration was slightly reduced in the elderly, which is in agreement with the results of DeVita and colleagues [10], whose elderly subjects walked at the same speed as the young subjects. The young and elderly subjects walked at a relatively slow speed (1.12 $m.s^{-1}$) compared with previous studies [4, 6], which was most probably due to the need to maintain the selected speed for a 20-minute period.

No differences in median MTC were detected between the young and elderly, a result which is consistent with previous studies [4, 6, 7]. This confirms the view that, despite an age-related reduction in the ability to recover from a trip [21], the elderly do not increase their overall safety margin for tripping. As hypothesised, the elderly exhibited kinematic differences at $time_{\mathrm{MTC}}$ including reduced stance hip extension and increased swing hip flexion in comparison with the young group. In the present study, hip joint angles were defined as neutral in the static anatomical trial and therefore these differences represent an age-related dynamic offset in sagittal plane hip kinematics in the direction of flexion. This dynamic offset in sagittal plane hip kinematics is consistent with previous findings [22] and is thought to be due to tightness or contractures of the hip flexor muscle group [22]. The elderly also exhibited 3.6 degrees less stance hip adduction than the young at $time_{\mathrm{MTC}}$. In isolation, MTC tends to increase by ~ 5 mm for each 1 degree reduction in stance hip adduction [9]. The finding that median MTC was similar for the age groups, while the elderly exhibited significantly less stance hip adduction suggests that the age groups employ different kinematic strategies to achieve a similar MTC.

As expected, MTC variability was greater in the elderly than the young, which is consistent with our hypothesis and the results of Begg and colleagues [7]. Given that the elderly do not exhibit a greater median MTC than the young, their greater MTC variability would result in a greater risk of the swing foot contacting the ground compared with the young. While other studies have identified age-related differences in stride duration [23] and step width [24] variability, these variables are somewhat removed from the mechanical cause of a fall. In contrast, an age-related increase in MTC variability, in the absence of an increase in median MTC, is indicative of an increased risk of the swing toe contacting the ground, and therefore has clear implications for trip related falls.

An unexpected finding was a lack of age-related differences in stance foot or joint angle variability at $time_{\mathrm{MTC}}$, despite a greater level of MTC variability in the elderly than the young. Moderate to large effect sizes point to an age-related increase in variability of stance hip flexion-extension, stance hip adduction-abduction, stance leg internal-external rotation angles and knee adduction-abduction angles of both legs, and indicate statistical significance may have been achieved with a larger sample. In addition, factors other than the magnitude of kinematic DOF variability may contribute to greater MTC variability in the elderly compared with the young. In some multiple kinematic DOF tasks requiring precision of a task specific motor output, fluctuations of individual kinematic DOFs covary to minimise motor output variability [25, 26]. Furthermore, this covariance is reduced in the elderly compared with the young [26], possible due to age-related decrements in proprioceptive acuity [27] and/or in the ability to modulate force [28]. It is possible that stride-to-stride fluctuations of the individual kinematic DOFs at $time_{\mathrm{MTC}}$ exhibit similar compensatory relationships to those observed in other precision motor tasks. If ageing adversely effects these relationships, an age-related increase in MTC variability could occur in the absence of marked increase in the variability of the individual kinematic DOFs involved in achieving MTC.

As hypothesised, joint angle variability at $time_{\mathrm{MTC}}$ was positively correlated with MTC variability for a number of kinematic DOFs. Variability of the sagittal plane swing leg ankle and knee joint angles were positively correlated with MTC variability for both the young and elderly groups. This corresponds with a sensitivity analysis performed by Winter [9] that demonstrated MTC was highly sensitive to fluctuations in swing ankle plantar-dorsiflexion and knee flexion-extension at $time_{\mathrm{MTC}}$. We also identified a positive relationship between stance ankle adduction-abduction variability and MTC variability in both age groups. Hof and colleagues [29] have proposed a dual-strategy control model of lateral balance during walking, in which gross control of lateral balance occurs via foot placement, while an ankle adduction-abduction strategy is used to make minor corrections throughout the stance phase. The model of Hof and colleagues [29] predicts that individuals who make greater lateral foot placement errors, would exhibit greater stride-to-stride variability in ankle adduction-abduction angles throughout stance in order to maintain lateral balance. This raises the question as to whether MTC variability is positively related to lateral instability, i.e., whether MTC variability is greater in individuals who are required to make greater corrections to lateral stability via an ankle adduction-abduction strategy. This question is worthy of further study, especially in the elderly population who, according to the results of the current study, have a strong positive relationship between ankle adduction-abduction variability and MTC variability.

In conclusion, although the young and elderly had similar median MTC values, the elderly exhibited a greater level of MTC variability than the young, which may increase their risk of a trip-related fall. We also identified age-related differences in median joint angle kinematics at $time_{\mathrm{MTC}}$ that were consistent with previous findings across the gait cycle. For both the young and elderly, joint angle variability at $time_{\mathrm{MTC}}$ was positively correlated with MTC variability for a number of kinematic DOFs, however no significant difference in joint angle variability was identified between the groups. It is possible that factors other the magnitude of joint angle variability at $time_{MTC}$ may contribute to the greater MTC variability of the elderly compared with the young. In conclusion, although the young and elderly had similar median MTC values, the elderly exhibited a greater level of MTC variability than the young, which may increase their risk of a trip-related fall.

Acknowledgements

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Table 1. Spatiotemporal and MTC gait variables of the young and elderly participants.

Spatiotemporal variables	Young	Elderly	Effect size
Gait velocity ($m.s^{-1}$)	1.12 ± 0.08	1.12 ± 0.09	0.00
Stride length (m)	1.27 ± 0.08	1.21 ± 0.08	0.75
Stride duration (s)	1.13 ± 0.05	1.09 ± 0.04	0.88 *
Stance (% of stride)	65.0 ± 1.5	63.6 ± 0.9	1.13 *
Swing (% of stride)	35.0 ± 1.5	36.5 ± 0.8	1.24 *
$time_{\mathrm{MTC}}$ (% of swing)	74.4 ± 3.2	73.4 ± 3.2	0.31
MTC median (mm)	14.9 ± 1.6	13.8 ± 2.1	0.59
MTC IQR (mm)	4.3 ± 0.9	5.3 ± 1.2	0.96 *

Mean ± SD; Positive effect sizes indicate greater values for elderly compared with young; * indicates significant difference compared with young, $P < 0.05$.

Table 2. Stance and swing leg joint angle medians for the young and elderly subjects at the time of minimum toe clearance.

	Stance leg			Swing leg		
Angle (deg)	Young	Elderly	Effect size	Young	Elderly	Effect size
Hip						
Flexion-extension	-11.6 ± 4.3	-2.6 ± 6.7 *	1.57	23.3 ± 3.8	30.5 ± 7.1 *	1.27
Adduction-abduction	6.3 ± 2.6	2.7 ± 3.7 *	-1.13	-0.8 ± 2.1	-2.7 ± 3.4	-0.64
Internal-external rotation	0.4 ± 7.8	-1.0 ± 5.3	-0.22	-3.3 ± 4.6	-4.4 ± 6.8	-0.17
Knee						
Flexion-extension	1.2 ± 5.6	3.0 ± 3.2	0.39	23.6 ± 5.7	24.1 ± 4.7	0.09
Adduction-abduction	-0.4 ± 3.8	2.0 ± 3.9	0.61	0.5 ± 2.6	3.6 ± 4.0 *	0.91
Internal-external rotation	-2.0 ± 6.3	-4.9 ± 2.9	-0.60	-6.1 ± 4.0	-5.8 ± 4.9	0.06
Ankle						
Plantar flexion-dorsiflexion	9.4 ± 5.1	10.9 ± 2.5	0.38	-6.2 ± 3.1	-4.7 ± 1.9	0.57
Inversion-eversion	-7.2 ± 4.0	-4.7 ± 3.2	0.69	-3.7 ± 2.8	-3.0 ± 0.9	0.35
Internal-external rotation	-3.7 ± 3.8	-2.8 ± 1.8	0.31	-3.0 ± 3.3	-3.4 ± 1.5	-0.14
Foot						
Flexion-extension	-0.2 ± 2.8	-0.7 ± 2.0	-0.41			
Adduction-abduction	-3.5 ± 4.0	-0.7 ± 5.8	0.57		—	
Internal-external rotation	-5.9 ± 4.1	-7.3 ± 5.3	-0.28			

Mean ± SD; Flexion, dorsiflexion, adduction and internal rotation joint angles are defined as positive while extension, plantar flexion, abduction and external rotation angles are defined as negative. Stance foot angles are global angles; Effect sizes are for age group (young vs. elderly), with positive values indicating greater values for elderly compared with young; * indicates significant difference compared with young, $P < 0.05$.

Table 3. Stance and swing leg joint angle inter-quartile ranges for the young and elderly subjects at the time of minimum toe clearance.

	Stance leg			Swing leg		
Angle (deg)	Young	Elderly	Effect size	Young	Elderly	Effect size
Hip						
Flexion-extension	1.7 ± 0.3	2.0 ± 0.5	0.85	1.6 ± 0.4	1.7 ± 0.4	0.19
Adduction-abduction	0.9 ± 0.2	1.0 ± 0.2	0.65	1.1 ± 0.2	1.1 ± 0.2	-0.15
Internal-external rotation	1.7 ± 0.4	1.8 ± 0.4	0.38	1.5 ± 0.3	1.5 ± 0.4	0.07
Knee						
Flexion-extension	1.9 ± 0.6	2.1 ± 0.5	0.40	2.0 ± 0.3	2.1 ± 0.3	0.39
Adduction-abduction	0.7 ± 0.2	0.9 ± 0.2	1.09	0.9 ± 0.2	1.0 ± 0.2	0.84
Internal-external rotation	2.7 ± 0.9	3.3 ± 0.9	0.66	1.8 ± 0.4	1.8 ± 0.2	0.02
Ankle						
Plantar flexion-dorsiflexion	1.2 ± 0.3	1.3 ± 0.2	0.18	1.3 ± 0.4	1.4 ± 0.3	0.19
Inversion-eversion	3.8 ± 0.7	3.5 ± 0.9	-0.41	1.1 ± 0.6	0.9 ± 0.2	-0.42
Internal-external rotation	2.1 ± 0.7	2.2 ± 0.8	0.07	1.1 ± 0.3	1.3 ± 0.2	0.46
Foot						
Flexion-extension	1.2 ± 0.5	1.1 ± 0.5	-0.05			
Adduction-abduction	1.4 ± 0.6	1.2 ± 0.6	-0.27	—		
Internal-external rotation	2.5 ± 1.0	2.4 ± 0.4	-0.20			

Mean ± SD; Effect sizes are for age group (young vs. elderly), with positive values indicating greater values for elderly compared with young; no significant age group differences were identified, $P < 0.05$.

Figure 1

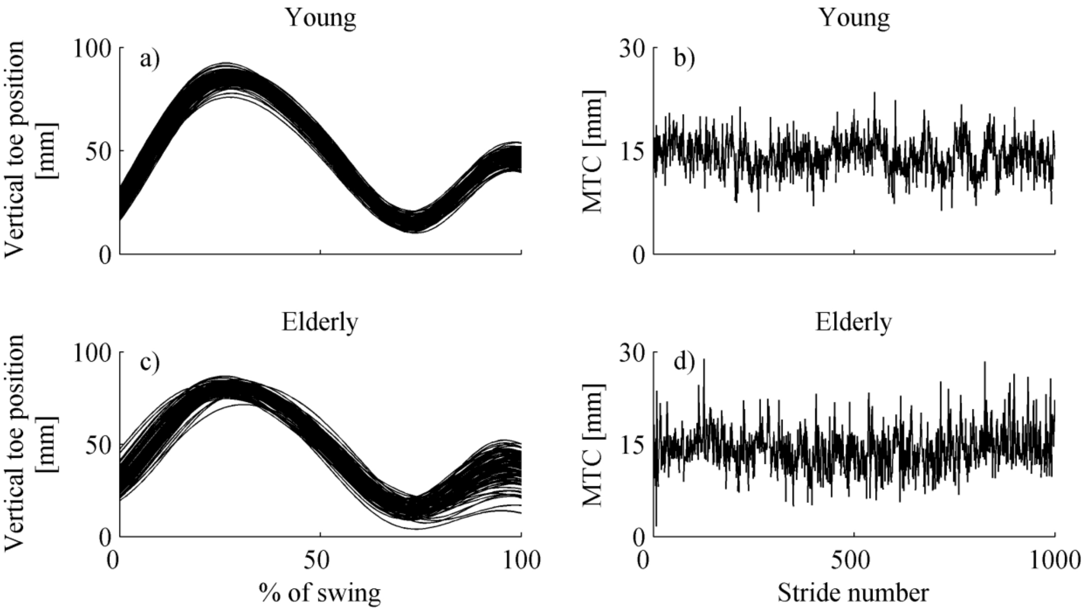
Young
Young
Elderly
Elderly
a)
b)
c)
d)
Vertical toe position [mm]
MTC [mm]
% of swing
Stride number
0
50
100
15
30
500
1000

Figure 2

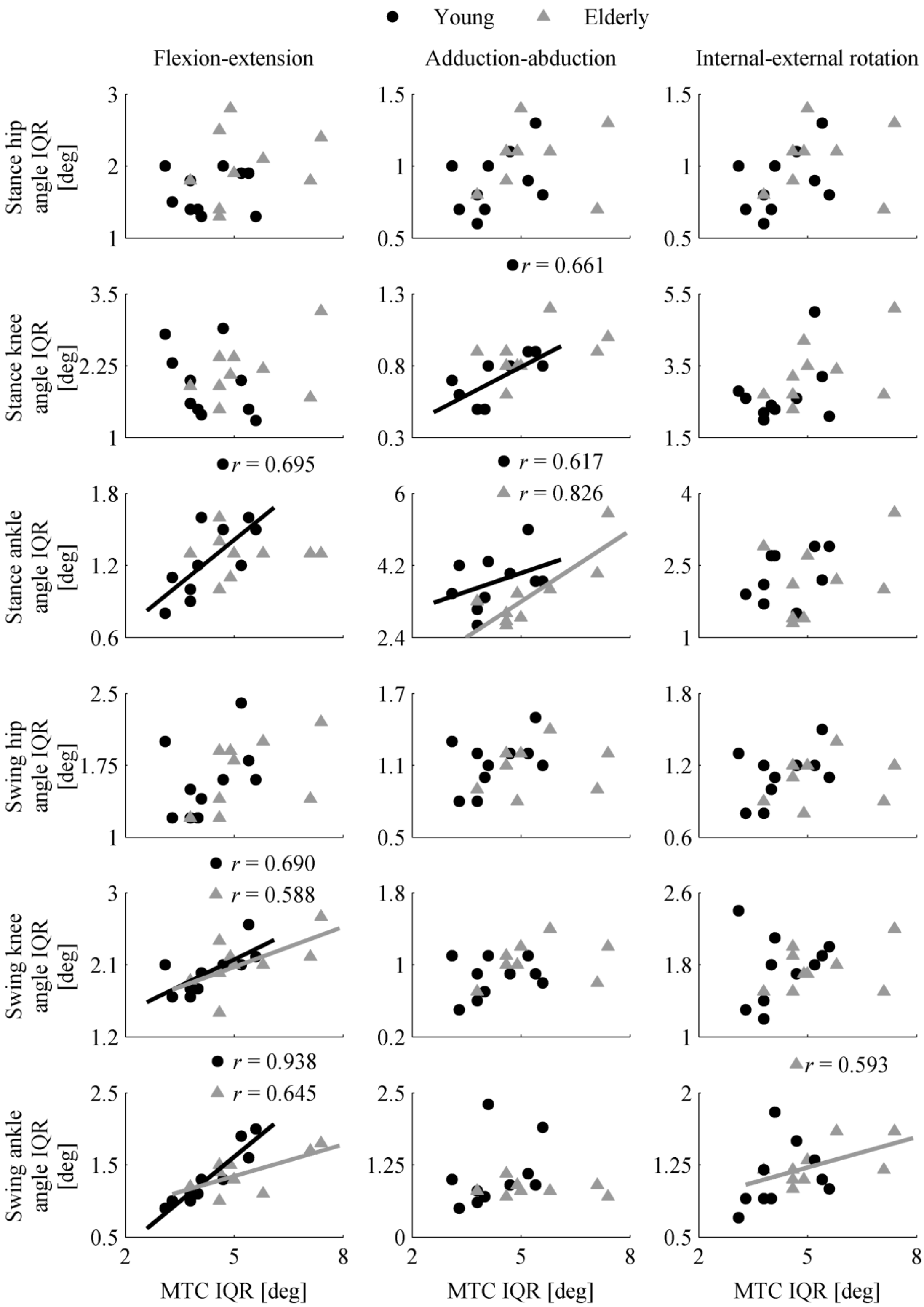
Young
Elderly
Flexion-extension
Adduction-abduction
Internal-external rotation
Stance hip angle IQR [deg]
Stance knee angle IQR [deg]
Stance ankle angle IQR [deg]
Swing hip angle IQR [deg]
Swing knee angle IQR [deg]
Swing ankle angle IQR [deg]
r = 0.661
r = 0.695
r = 0.617
r = 0.826
r = 0.690
r = 0.588
r = 0.938
r = 0.645
r = 0.593
MTC IQR [deg]
MTC IQR [deg]
MTC IQR [deg]

Figure captions

Figure 1. Representative toe clearance for the young and elderly. Vertical swing toe position throughout the swing phase for 100 cycles for representative young (a) and elderly (c) subjects. Minimum toe clearance (MTC) for 1000 cycles for representative young (b), and elderly (d) subjects.

Figure 2. Scatter plots of minimum toe clearance (MTC) inter-quartile ranges (IQRs) vs. joint angle IQRs at the time of minimum toe clearance for the young and elderly subjects. Rows indicate the joint while columns indicate the axis of rotation. Regression lines and r values are presented only for significant age group correlations (P<0.05, see text for actual P-values), with black lines indicating a significant correlation within the young group and grey lines indicating a significant correlation within the elderly group.