ARTICLE IN PRESS

Journal of Biomechanics ∎ (∎∎∎∎) ∎∎∎–∎∎∎

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Estimation of minimum ground clearance (MGC) using body-worn inertial sensors

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ARTICLE INFO

Article history:
Accepted 31 January 2011

Keywords:
Inertial sensor
Minimum ground clearance
Falls risk

ABSTRACT

Objective assessment of balance and mobility in elderly populations using body-worn sensors has recently become a prevalent theme in falls-related research. Recent research by the authors identified mean absolute-valued vertical angular velocity measured using shank mounted inertial sensors during a timed-up-and-go test as having a strong association with falls history in a group of elderly adults. This study aimed to investigate the clinical relevance of this parameter by exploring the relationship between it and minimum ground clearance (MGC) measured with an optical motion capture system. MGC is an important variable when considering trip-related falls risk. This paper also presents a method of estimating properties of MGC during walking, across a range of speeds and gait patterns, using body-worn inertial sensors. We found that mean MGC and coefficient of variation (CV) MGC are correlated with mean absolute-valued vertical angular velocity and acceleration as measured by shank or foot mounted inertial sensors. Regression models generated using inertial sensor derived variables were used to robustly estimate the mean MGC and CV MGC measured by an optical marker-tracking system. Foot-mounted sensors were found to yield slightly better results than sensors on the shank. Different walking speeds and gait patterns were not found to influence the accuracy of the models. We conclude that these findings have the potential to evaluate a walking trial using body-worn inertial sensors, which could then be used to identify individuals with increased risk of unprovoked collisions with the ground during locomotion.

1. Introduction

Objective assessment of balance and mobility in elderly populations using body-worn sensors has recently become a prevalent theme in falls-related research. The potential for body-worn sensors as a low-cost, light-weight tool for extra-laboratory gait monitoring is well recognised. However, the large amount of information that these sensors can generate, and the potential relationship between this information and clinically meaningful parameters has not yet been fully exploited. Recent research by the authors found that mean absolute-valued vertical angular velocity measured by a shank mounted inertial sensor during a timed-up-and-go test was strongly associated with falls history in a cohort of 349 community dwelling elderly adults (Greene et al., 2010). This parameter was measured over the entire walking task, which was comprised of several gait cycles, resulting in a single value. This finding revealed a simple parameter that could potentially be used as part of a falls risk assessment tool, through objective assessment of a walking trial. However, the causative relationship between mean absolute-valued vertical angular velocity and falls history was not evident from the data.

The task of walking is a complex motor control challenge, where the human neuromuscular system is required to interact with the environment while maintaining balance during forward momentum. One of the most important outputs of the locomotor system is safe foot trajectory with respect to the ground. Winter demonstrated that foot-trajectory during the swing phase of gait is very sensitive to small angular changes at six other joints within both stance and swing limbs, providing evidence that it is a very precise end-point control task (Winter, 1992). Disturbance of this end-point control may lead to unprovoked collision with the ground in affected populations.

The local minimum distance between the foot/shoe and the ground during swing can be referred to as minimum ground clearance (MGC). At this critical instant, the foot is at or near its maximum velocity; the body is in single limb stance with the centre of mass outside the base of support and in the direction of

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doi:10.1016/j.jbiomech.2011.01.034

progression (Winter, 1992). MGC typically measures approximately 10–15 mm (Winter, 1992; Karst et al., 1999; Begg et al., 2007), so a small positional error at this point could result in a collision with the ground, with deleterious effects. A recent review by Goble et al. (2009) on proprioception in the elderly found that studies comparing young and elderly subjects have, almost unanimously, indicated a significant deterioration of joint or segment position sense with age. Defective joint position sense has also been reported in patients with chronic ankle instability (Nakasa et al., 2008). Therefore, MGC could be regarded as a critical variable of interest when considering trip-related falls in the elderly or indeed common musculoskeletal complaints such as repeated ankle sprains.

Previous investigations have used body-worn sensors to measure foot-trajectory during swing. Lai et al. (2008) devised a prototype device for measuring MGC by embedding inertial sensors into a running shoe. However, foot displacement was computed through double-integration of acceleration data, which lead to cumulative errors over longer measurement durations. In a recent study by Mariani et al. (2010), the authors used a de-drifted double integration technique to calculate maximal foot height, where the initial conditions of sensor position and orientation were updated for each gait cycle. Minimum foot height during swing (i.e. MGC) was not addressed in this study.

The purpose of this study was to investigate the clinical significance of absolute-valued vertical angular velocity by exploring the relationship between it and the mean and coefficient of variation (CV) of MGC (referred to hereafter as mean MGC and CV MGC, respectively) measured using an optical motion capture system over an entire walking trial. It was envisioned that the results of this study would substantiate the potential use of this parameter as part of a falls risk assessment tool, through macro, objective assessment of a walking trial. A second aim of the study was to find surrogate measures for mean and CV MGC using statistical models based on parameters derived from inertial sensors. We examined variables derived from both gyroscopes and accelerometers for estimation of MGC properties. We also investigated two different sensor locations – foot and shank – to determine the more suitable location. The participants walked at four different speeds, which included a mimicked shuffling gait pattern to see its effect on the robustness of the model.

2. Methods

2.1. Data acquisition

The gait of nine healthy subjects (8M, 1F, mean age: 29.7 ± 3.5) was measured simultaneously using two gait measurement technologies: wireless inertial sensors, and an optical motion capture system. Informed consent was obtained from each participant once the purpose and requirements of the study were explained to them.

Data were recorded whilst each subject performed four walks at four different self-selected speeds – shuffle, slow, normal, and fast – along a 15 m walkway in a motion analysis laboratory. In all, 16 walking trials were completed per subject (144 trials in all). All speeds were self-selected and not controlled to minimise the effect of the experimental protocol on the subjects' gait.

2.2. Marker-based data acquisition

Reference kinematic data were acquired using a CODA optical motion analysis system (http://www.codamotion.com, Charnwood Dynamics Ltd., Leicestershire, UK). Two CODA infrared light-emitting diode markers were placed on the left and right foot. Markers were positioned on the inferior lateral aspect of the heel, and the lateral aspect of the fifth metatarsal head, on the exterior of the participants' training shoes. The optical kinematic data were collected at a sampling rate of 200 Hz. Kinematic data were analysed using the CODAmotion analysis software.

2.3. Minimum ground clearance calculation

The method used here for measurement of MGC calculates the minimum vertical displacement during the swing phase of a marker positioned on the lateral aspect of the fifth metatarsal head (Dingwell et al., 1999; Osaki et al., 2007). While this method may not measure absolute MGC between the foot/shoe and the ground, it is an appropriate method for the purpose of this study, where the relationship between MGC and inertial sensor-based parameters is being explored. MGC values reported here are not therefore absolute values, and are not corrected for shoe height. The mean and CV MGC were calculated for each individual walking trial. An average of 104 MGC points per participant were used in the analysis.

2.4. Inertial sensor data acquisition

Inertial sensor data were acquired using four SHIMMER wireless sensors (Burns et al., 2010), one each attached to the mid-foot and shank of the left and right leg, as shown in Fig. 1. Sensors were attached to the skin using double sided tape and secured with additional tape. Each sensor contained both a tri-axial accelerometer and a tri-axial gyroscope, sampling each axis at 102.4 Hz. Data were acquired from the tri-axial gyroscopes and accelerometers using a custom developed application (http://www.biomobius.org) (Burns et al., 2010). All post-processing and analysis were carried out off-line using the MATLAB® (http://www.mathworks.com/ (Natick, VA, USA)) programming environment. The raw

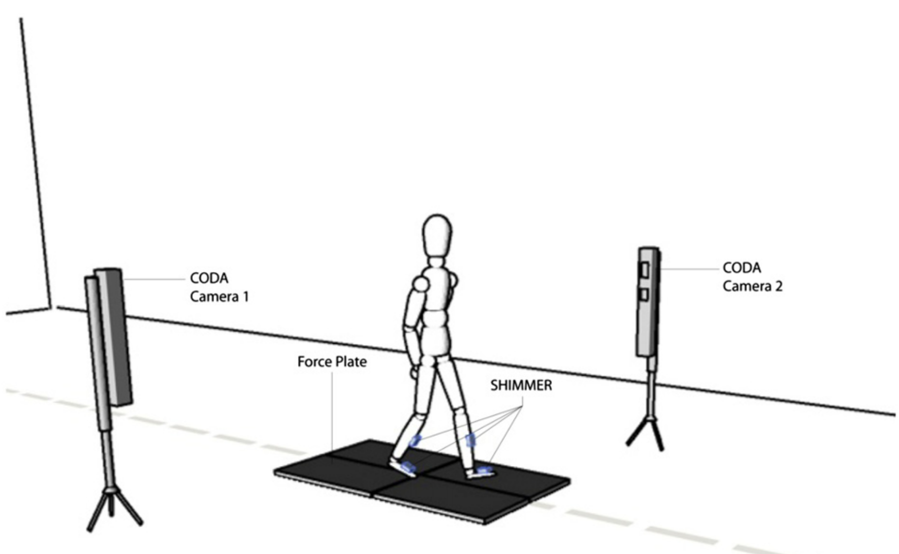

Fig. 1. Experimental setup for data-acquisition from optical motion capture system and wireless inertial sensors.

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gyroscope and accelerometer data were calibrated to derive the angular velocity and acceleration vectors with respect to the sensor unit coordinate axis. A standard calibration procedure (Ferraris et al., 1995) was used to calibrate all gyroscopes and accelerometers used in the study. Before further processing, the raw signal from each sensor was low pass filtered with zero-phase 5th order Butterworth filter with a 10 Hz corner frequency.

The inertial sensor data acquisition system and the optical motion capture systems were synchronized using a dedicated trigger output from the CODA system, which was activated at the initiation and deactivated at the conclusion of a data-capture. This trigger signal was sampled at 102.4 Hz by a separate SHIMMER device and transmitted wirelessly via bluetooth to the acquisition software. Data from the synchronization and kinematic SHIMMER devices were simultaneously recorded within BioMOBIUS. Data acquired by the motion capture system were down-sampled to the nearest sensor sample.

2.5. Inertial sensor parameters

The angular velocity of the shank was measured about an axis perpendicular to the sagittal plane and relates to the orientation change of the segment in the sagittal plane. Acceleration was measured in a direction parallel to gravity. A number of measures were calculated from the filtered angular velocity and vertical acceleration signals measured over an entire walking trial with body-worn sensors mounted on the foot and shank of each leg for each subject, listed below:

- Mean absolute value
- Max
- Min
- Range

Derived variables were used to generate statistical models for estimating the mean and variability of the MGC over each walking trial for each subject. Each derived variable is calculated over the synchronized portion of the vertical angular velocity and acceleration signals for each of the 16 walking trials for each subject. This yielded 16 data points per leg, for each walking trial per subject.

2.6. Statistical analysis

Regression models were used to determine if the gyroscope and accelerometer derived variables could be used to generate a robust estimate of mean and variability of MGC over a given walking trial. The mean and CV of the MGC points for a given walking trial were used as response variables for these models. Predictors were identified using the results from correlation analysis. In order to determine if there was a linear relationship between each of the derived variables and the mean and CV MGC, Pearson's correlation coefficient was used. Data for left and right legs were analysed separately initially, avoiding any assumptions of gait symmetry.

Regression models for both mean MGC and CV MGC were then generated using subject number (1–9) and leg (left or right) as categorical factors using dummy variables. Linear, quadratic, and stepwise quadratic regression models were generated. Separate models were generated for foot and shank sensors. Finally, regression models that combined both limbs and all subjects' data points across all speeds and gait patterns – fast, medium, slow, and shuffle – were generated.

3. Results

3.1. Correlation analysis

Comparison of the vertical displacement trace of the marker placed on the 5th metatarsal head obtained from the optical motion capture system to the vertical angular velocity and acceleration signals obtained from the foot and shank mounted inertial sensors suggested a concordance between the two (see Fig. 2).

The Pearson correlation coefficient between the mean and CV MGC and each vertical angular velocity and acceleration derived variables was calculated for the foot and shank for each leg. The variables that best correlated with mean MGC and CV MGC were mean absolute-valued vertical angular velocity and acceleration. The largest mean correlation coefficient with mean MGC (calculated across all subjects) was found between the mean absolute-valued vertical angular velocity derived from foot mounted inertial sensors (left, $r=0.77$; right, $r=0.72$). Similarly, the mean absolute-valued vertical acceleration from foot mounted inertial sensors showed good correlation with the mean MGC (left, $r=0.74$; right, $r=0.70$). The mean correlation of mean absolute-valued vertical angular velocity and acceleration signals for the shank mounted sensors was not as strong as for the foot-mounted sensor (gyroscope: left, $r=0.73$; right, $r=0.63$, accelerometer: left, $r=0.73$; right, $r=0.68$). Similarly, correlation relationships were

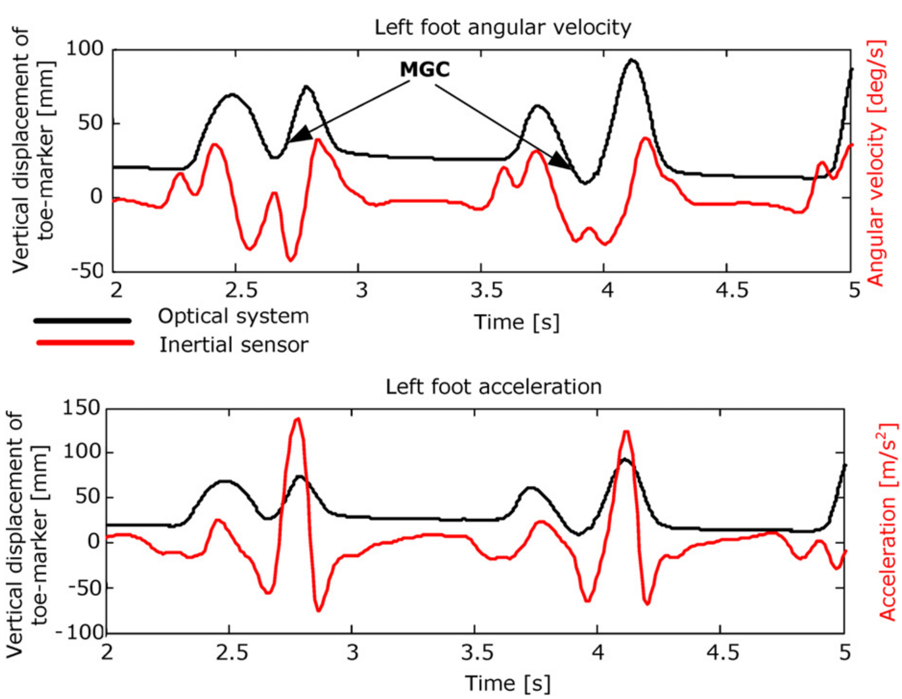

Fig. 2. Vertical displacement of toe marker measured by optical motion capture system showing concordance with angular velocity and acceleration signals derived from inertial sensor mounted on left foot. Angular velocity of the shank was measured about an axis perpendicular to the sagittal plane and relates to the orientation change of the segment in the sagittal plane. Acceleration was measured in the direction parallel to gravity. The MGC points on the optical system trace are marked.

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Table 1
Results for regression models for estimating mean and coefficient of variability of MGC from foot and shank mounted tri-axial inertial sensors. Results for linear, quadratic, and stepwise quadratic regression models are given.

Model	Mean MGC						CV MGC					
	Foot			Shank			Foot			Shank		
	R^2	F	Error	R^2	F	Error	R^2	F	Error	R^2	F	Error
Linear	0.66	43.26	35.86	0.60	34.26	41.53	0.37	8.03	62.19	0.33	7.25	65.57
Quadratic	0.85	33.63	17.34	0.82	25.87	21.58	0.76	2.67	48.50	0.70	2.75	50.75
Stepwise quadratic	0.83	64.36	17.77	0.80	46.58	21.37	0.54	19.36	45.15	0.43	18.28	54.22

Table 2
Parameters included in the regression models used to estimate the mean and CV MGC for each walking trial for each subject.

Mean MGC	CV MGC
Mean abs vert. ang. vel.	Mean abs vert. ang. vel.
Mean abs accel.	Max vert. ang. vel.
	Min vert. ang. vel.
	Range vert. ang. vel.
	Vert. ang. vel. at mid-swing
	Mean abs accel.
	Max vert. accel.
	Min vert. accel.
	Range vert. accel.
	Vert. accel. at mid-swing

analysed between CV MGC and mean absolute values of vertical angular velocity and acceleration for the foot mounted sensors (gyroscope: left, $r=-0.59$; right, $r=-0.45$, accelerometer: left, $r=-0.56$; right, $r=-0.48$) as well as for the shank mounted sensors (gyroscope: left, $r=-0.58$; right, $r=-0.48$, accelerometer: left, $r=-0.55$; right, $r=-0.48$).

3.2. Regression analysis

A quadratic regression model relating mean MGC to mean absolute value vertical angular velocity and mean absolute-valued vertical acceleration from a foot mounted gyroscope achieved R^2 of 0.85, $p<0.001$. Similarly a quadratic regression model of CV MGC resulted in an R^2 value of 0.76, $p<0.001$. All data from all 9 subjects, i.e. both limbs and all gait speeds and patterns, were combined in these models. Walking speeds taken across all subjects were as follows: fast, 1.56 m/s (5.61 km/h); normal, 1.10 m/s (3.95 km/h); slow, 0.65 m/s (2.36 km/h); and shuffle, 0.48 m/s (1.71 km/h).

Table 1 provides results for regression models of mean and CV MGC for the foot and shank mounted using inertial sensor derived parameters. The parameters included in the regression models for estimating mean and CV MGC are listed in Table 2.

Figs. 3 and 4 compare the predicted MGC output from the quadratic regression models for foot and shank mounted sensors against the reference mean MGC and CV MGC data obtained from the optical motion capture system.

4. Discussion

The primary aim of this study was to investigate the clinical relevance of mean absolute-valued vertical angular velocity measured across an entire walking trial using an inertial sensor attached to the shank following a previous study that identified this parameter as having a strong association with falls history among 349 community dwelling elderly adults (Greene et al., 2010). Our results show a correlation between this parameter and mean and CV MGC, measured using an optical motion capture system. This finding represents a step towards exploiting novel parameters generated by inertial sensors for meaningful clinical applications. Objective evaluation of an entire walking trial using this parameter could potentially be used as part of a quantitative falls risk assessment. The evidence provided here that this parameters relates to properties of MGC supports the clinical value of such an approach.

Despite the widely accepted association between MGC and trip-related falls, there is limited evidence in the literature that directly relates MGC to falls risk. Khandoker et al. (2008) put forward the most convincing evidence, where an array of non-linear variability measures were applied to MGC data, three of which were found to be potential markers that could reliably identify fall risk subjects from healthy elderly subjects The present study shows that mean absolute-valued vertical angular velocity measured from the shank – a parameter previously shown to discriminate between fallers and non-fallers – is correlated with MGC. This finding offers additional evidence to support the assertion that MGC is related to falls risk.

A second aim of this study was to propose a method for using body-worn inertial sensors to estimate the mean and variability of MGC for a given walking trial, at a variety of self-selected walking speeds. Previous research has found that elderly men exhibit greater MGC variability than young men, in the absence of any age-related differences in average MGC values (Mills et al., 2008). This suggests that variability of MGC may be a more important parameter in evaluating increased risk of tripping in the elderly. CV MGC was selected as a measure of MGC variability. Both parameters were found to correlate with mean absolute-valued vertical angular velocity and acceleration as measured by shank or foot mounted inertial sensors. Regression models were developed using these variables to estimate mean MGC and CV MGC measured by an optical marker-tracking system. Figs. 3 and 4 illustrates that quadratic regression models robustly predicted true mean and CV MGC values. Foot-mounted sensors were found to yield slightly better results than sensors on the shank.

The relationships between MGC and the inertial sensor derived variables held true across a range of walking speeds, and gaits. We asked our participants to mimic a shuffling gait in order to test the predictive value of our models in non-normal gait patterns. This instruction was interpreted in many different ways, varying from participants barely lifting their feet off the ground, to taking very short, rapid steps, with greater ground clearance. Our models proved robust enough to deal with this range of artificially aberrant gait patterns.

This is a proof of concept study that proposes a novel method for quantitatively evaluating an entire walking trial using body-worn inertial sensors. We have provided evidence that endorses this method as a clinically meaningful approach by relating it to MGC. We conclude that robust estimation of mean and CV MGC from shank or foot mounted inertial sensors (containing tri-axial gyroscopes and tri-axial accelerometers) using regression models,

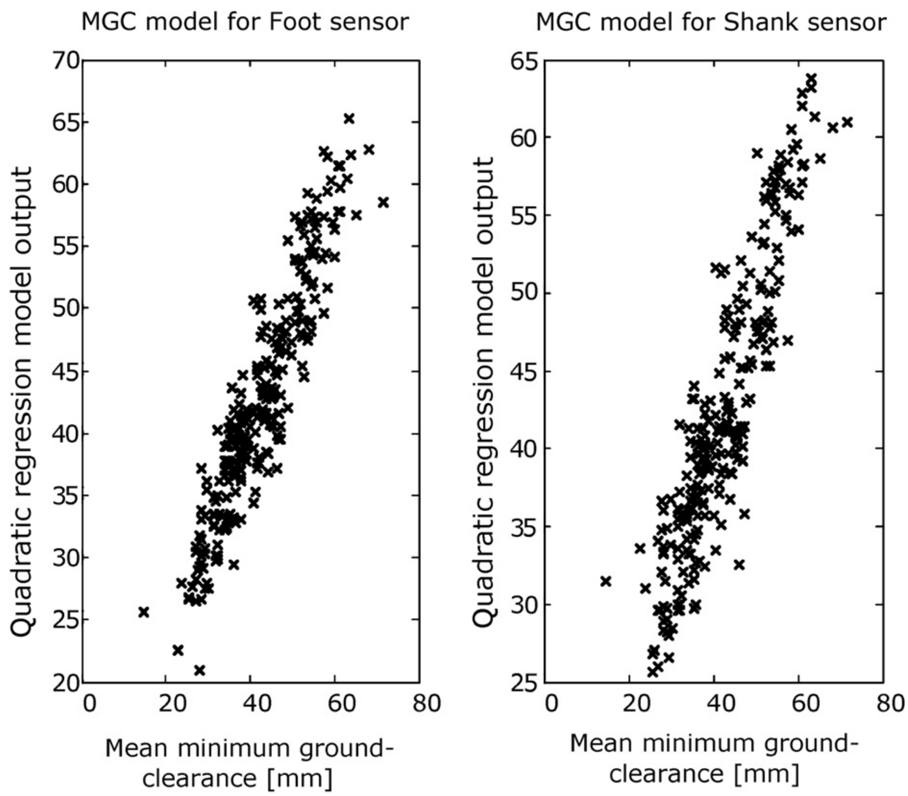

Fig. 3. Comparison of output from quadratic regression of mean MGC to mean MGC values derived from optical motion capture system for foot and shank mounted sensors.

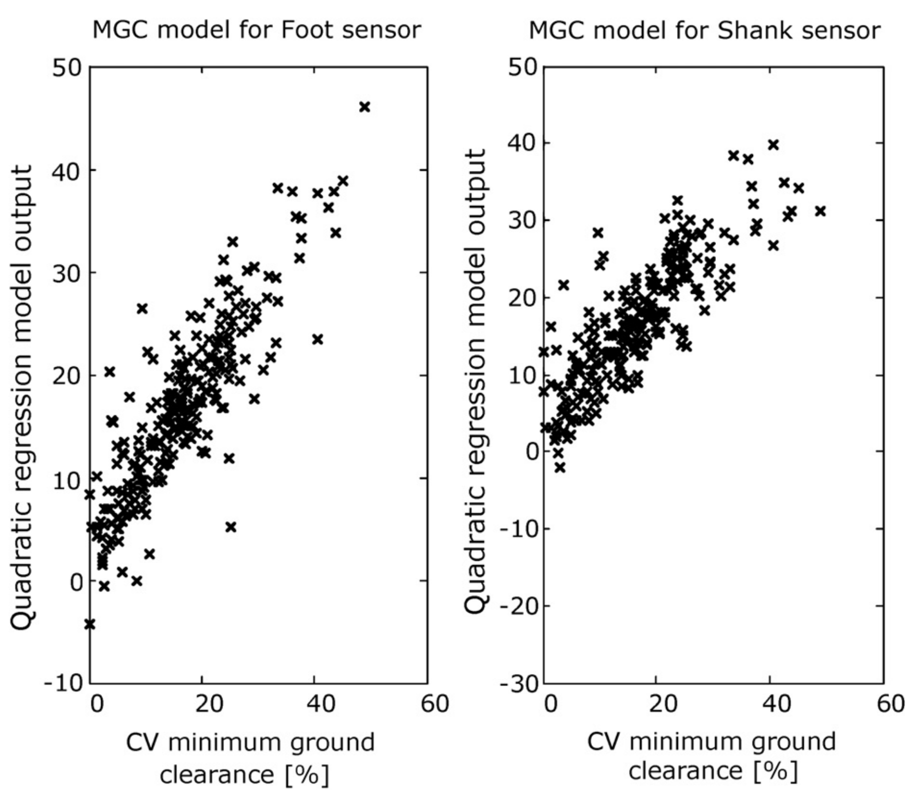

Fig. 4. Comparison of output from quadratic regression model of CV MGC to CV MGC values derived from optical system for foot and shank mounted sensors for all data points.

is possible. These findings have the potential to enable accurate, extra-laboratory monitoring of MGC across an entire walking trial that could then be used in identifying individuals with increased risk of unprovoked collisions with the ground during locomotion. Future work will seek to validate these findings in an elderly population.

Conflict of interest statement

Funding and SHIMMER hardware were provided by the Intel Corporation and the TRIL centre.

Acknowledgements

This research was completed as part of a wider programme of research within the TRIL Centre, (Technology Research for Independent Living). The TRIL Centre is a multi-disciplinary research centre, bringing together researchers from UCD, TCD, NUIG & Intel, funded by Intel, IDA Ireland and GE Healthcare. www.trilcentre.org. The authors would like to thank Ms. Jessica Wegelin for her help with graphically detailing the experimental layout.

References

Begg, R., Best, R., Dell'oro, L., et al., 2007. Minimum foot clearance during walking: strategies for the minimisation of trip-related falls. Gait Posture 25 (2), 191–198.

Burns, A., Greene, B.R., Mcgrath, M.J., et al., 2010. SHIMMER™—a wireless sensor platform for non-invasive biomedical research. IEEE Sensors Journal, IEEE Sensors, 101527–101534.

Dingwell, J.B., Ulbrecht, J.S., Boch, J., et al., 1999. Neuropathic gait shows only trends towards increased variability of sagittal plane kinematics during treadmill locomotion. Gait & Posture 10 (1), 21–29.

Ferraris, F., Grimaldi, U., Parvis, M., 1995. Procedure for effortless in-field calibration of three-axis rate gyros and accelerometers. Sensor and Materials 7 (5), 311–330.

Goble, D.J., Coxon, J.P., Wenderoth, N., et al., 2009. Proprioceptive sensibility in the elderly: degeneration, functional consequences and plastic-adaptive processes. Neuroscience & Biobehavioral Reviews 33 (3), 271–278.

Greene, B.R., O'donovan, A., Romero-Ortuno, R., et al., 2010. Quantitative falls risk assessment using the timed up and go test. IEEE Transactions Biomedical Engineering, 572918–572926.

Karst, G.M., Hageman, P.A., Jones, T.F., et al., 1999. Reliability of foot trajectory measures within and between testing sessions. Journal of Gerontology A: Biological Sciences and Medical Sciences 54 (7), M343–M347.

Khandoker, A.H., Taylor, S.B., Karmakar, C.K., et al., 2008. Investigating scale invariant dynamics in minimum toe clearance variability of the young and elderly during treadmill walking. IEEE Transactions on Neural Systems and Rehabilitation Engineering 16 (4), 380–389.

Lai, D.T., Charry, E., Begg, R., et al., 2008. A prototype wireless inertial-sensing device for measuring toe clearance. Conference Proceedings IEEE Engineering in Medicine and Biology Society 20084899-902.

Mariani, B., Hoskovec, C., Rochat, S., et al., 2010. 3D gait assessment in young and elderly subjects using foot-worn inertial sensors. Journal of Biomechanics 43 (15), 2999–3006.

Mills, P.M., Barrett, R.S., Morrison, S., 2008. Toe clearance variability during walking in young and elderly men. Gait Posture 28 (1), 101–107.

Nakasa, T., Fukuhara, K., Adachi, N., et al., 2008. The deficit of joint position sense in the chronic unstable ankle as measured by inversion angle replication error. Archives of Orthopaedic and Trauma Surgery 128 (5), 445–449.

Osaki, Y., Kunin, M., Cohen, B., et al., 2007. Three-dimensional kinematics and dynamics of the foot during walking: a model of central control mechanisms. Experimental Brain Research 176 (3), 476–496.

Winter, D.A., 1992. Foot trajectory in human gait: a precise and multifactorial motor control task. Physical Therapy 72 (1), 45–53 discussion 54-6.