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Reliability study of tibialis posterior and selected leg muscle EMG and multi-segment foot kinematics in rheumatoid arthritis associated pes planovalgus

Ruth Barn *, Daniel Rafferty, Deborah E. Turner, James Woodburn

Institute for Applied Health Research, School of Health and Life Sciences, Glasgow Caledonian University, Glasgow, United Kingdom

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ABSTRACT

Objective: To determine within- and between-day reliability characteristics of electromyographic (EMG) activity patterns of selected lower leg muscles and kinematic variables in patients with rheumatoid arthritis (RA) and pes planovalgus.

Methods: Five patients with RA underwent gait analysis barefoot and shod on two occasions 1 week apart. Fine-wire (tibialis posterior [TP]) and surface EMG for selected muscles and 3D kinematics using a multi-segmented foot model was undertaken barefoot and shod. Reliability of pre-determined variables including EMG activity patterns and inter-segment kinematics were analysed using coefficients of multiple correlation, intraclass correlation coefficients (ICC) and the standard error of the measurement (SEM).

Results: Muscle activation patterns within- and between-day ranged from fair-to-good to excellent in both conditions. Discrete temporal and amplitude variables were highly variable across all muscle groups in both conditions but particularly poor for TP and peroneus longus. SEMs ranged from 1% to 9% of stance and 4% to 27% of maximum voluntary contraction; in most cases the 95% confidence interval crossed zero. Excellent within-day reliability was found for the inter-segment kinematics in both conditions. Between-day reliability ranged from fair-to-good to excellent for kinematic variables and all ICCs were excellent; the SEM ranged from 0.60° to 1.99°.

Conclusion: Multi-segmented foot kinematics can be reliably measured in RA patients with pes planovalgus. Serial measurement of discrete variables for TP and other selected leg muscles via EMG is not supported from the findings in this cohort of RA patients. Caution should be exercised when EMG measurements are considered to study disease progression or intervention effects.

1. Background

Rheumatoid arthritis (RA) is an inflammatory polyarthritis with a variable course and prognosis. The prevalence is approximately 1% and affects females more commonly than males (3:1) [1]. Upwards of 80% of patients report foot problems during the course of the disease [2]. Acquired pes planovalgus (PPV) with associated involvement of the tibialis posterior (TP) tendon is common [3]. Pes planovalgus is a complex multi-planar deformity characterised by valgus alignment of the rearfoot with corresponding midfoot collapse and forefoot abduction [4]. The cause is unknown but inflammatory damage in the peri-talar complex, tibialis posterior (TP) tendinopathy and altered foot mechanics frequently co-exist in published cohorts [3–6].

The contribution of TP muscle dysfunction to the deformity is unknown. Currently only one study has investigated electromyographic (EMG) activity of TP in RA [7]. Here the magnitude of activity of TP in stance was increased in patients with valgus rearfoot alignment compared to those without [7]. These findings have been replicated in non-RA cohorts with TP tendon dysfunction and flat foot [8,9]. Despite preliminary evidence regarding differences in EMG activity of TP with varying foot posture; there is minimal evidence in terms of how these features respond to intervention or how they behave over time in a chronic inflammatory joint disease such as RA. This merits further study.

The consequence of TP dysfunction can also be indirectly assessed through structural and functional changes, for example by 3D kinematics of the peri-talar complex via multi-segmented foot models. Previous investigations into PPV in RA have employed models with 3-segments or less [4,6,10]. Here we aim to develop and understand the reliability characteristics of an extended

* Corresponding author at: Department of Psychology and Allied Health Sciences, School of Health and Life Sciences, Glasgow Caledonian University, Glasgow G4 0BA, United Kingdom. Tel.: +44 141 331 3869.
E-mail addresses: ruth.barn@gcu.ac.uk (R. Barn), d.rafferty@gcu.ac.uk (D. Rafferty), debbie.turner@gcu.ac.uk (D.E. Turner), jim.woodburn@gcu.ac.uk (J. Woodburn).

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multi-segmented foot model which includes a midfoot segment combined with fine-wire EMG to TP and surface EMG to selected lower limb muscles applied in patients with RA and PPV. In particular, the motivation to investigate between-day reliability is a prerequisite for accurately quantifying progressive change in muscle and joint function associated with chronic inflammatory disease, and as a technique to assess, monitor and evaluate function and outcome for non-pharmacological, pharmacological and surgical interventions.

2. Methods

2.1. Patients

Patients were recruited from outpatient clinics at Glasgow Royal Infirmary and Gartnavel General Hospital, Glasgow UK. Patients were eligible for inclusion if they had a confirmed diagnosis of RA based on the 1987ACR criteria [11] and passively correctable PPV [4]. Patients were excluded if they had a history of a significant neurological or musculoskeletal condition affecting the lower limb in terms of gait or muscle function or if they were taking anti-coagulant medication. The study was conducted in accordance with the Declaration of Helsinki and ethical approval was obtained from the West of Scotland Local Research Ethics Committee (09/S0704/44) and NHS Greater Glasgow and Clyde Research and Development (GN09RH373).

2.2. Demographic, disease and clinical assessment

Participant age, gender and disease duration were recorded. To account for changes in foot function, symptoms or treatment status as a result of the disease process, a core set of clinical variables were recorded. These included: a tender and swollen foot joint count undertaken by a single clinician (RB); foot posture using the Structural Index [12]; foot related impairment and disability using the Foot Impact Scale (FIS) for RA and global disability using the Health Assessment Questionnaire (HAQ). Visual analogue scales (100 mm VAS) were used to record foot pain, general health and arthritis pain and change in therapy between time points was recorded.

2.3. Biomechanical analysis

Twenty spherical reflective markers (5 and 10 mm diameter) were located on anatomical landmarks during static calibration. This was reduced to 13 and 11 tracking markers for the barefoot and shod conditions respectively (detailed description of the model is contained within Supplementary material). A 12 camera 120 Hz motion analysis system (Qualisys Oqus, Gothenburg, Sweden) was used to track the motion of the markers during gait trials. Visual 3D software (C-Motion, Inc., Rockville, MD, USA) was used to build the multi segmented foot model. The model comprised five segments for the barefoot condition comprising the whole foot, shank, rearfoot, midfoot and forefoot. The midfoot segment could not be recorded in shoe so the model was reduced to four segments in the shod condition. Five trials were recorded and the ensemble average created for each condition on two occasions 1 week apart. The limb selected for analysis was based on clinical severity of foot posture.

Previous work in inflammatory joint disease [10,13,14] has informed the selection of a core set of discrete kinematic variables which best represent foot dysfunction. These include initial foot contact angle and terminal stance plantarflexion angle, peak rearfoot eversion angle, peak midfoot inversion angle, lowest navicular height (mm), peak forefoot dorsiflexion angle and peak forefoot abduction angle.

A shoe (Flextop Diabetic Shoe, Reed Medical Ltd., UK) was modified for the shod trials. Windows were cut to allow the tracking of markers for the rear and forefoot segments and to record vertical height of the navicular. The shoe was selected for the soft leather upper and flexible vamp in order to accommodate forefoot deformity. Velcro straps were stitched to the superior aspect of the heel counter in order to maintain stability during walking (Supplementary file).

2.4. EMG analysis

In order to avoid undertaking an invasive procedure on subjects potentially at risk of infection, intramuscular EMG was restricted to the TP muscle because of its inaccessibility via surface electrodes. Tibialis anterior, soleus, peroneus longus and medial gastrocnemius EMG signals were recorded using Trigno™ wireless surface electrodes (Delsys Inc., Boston, USA) following the SENIAM recommendations [15]. Surface electrodes had a single differential configuration, inter-electrode distance of 10 mm, 4-bar formation, bandwidth of 20–450 Hz and 99.9% silver contact material.

Intramuscular EMG of TP was undertaken using bi-polar stainless steel nylon coated fine wire electrodes (0.051 mm diameter) inserted via 50 mm length 15 gauge needles (Motion Lab Systems Inc., LA, USA). The raw signal was passed through a differential amplifier with a gain of 1000 and sampled at a frequency of 2400 Hz. Electrodes were inserted under ultrasound guidance (Esaote Mylab 70) using a 13–4 MHz linear array transducer via the posterior-medial approach at 50% of the distance between the medial malleolus and the medial joint line of the knee [16]. Placement of the electrode was verified by checking the signal while applying manual resistance in the direction of dorsiflexion and eversion while participants plantarflexed and inverted; the signal was also checked when participants flexed their toes to ensure the electrode was not placed inadvertently in the flexor digitorum longus muscle. Walking time over 5.5 m was recorded using timing gates (Brower Timing Systems, Utah, USA) and trials exceeding ±5% of the self-selected speed were excluded. Walking speed on the second visit was matched to the initial testing session ±5%.

2.5. Data processing

All EMG signals were high pass filtered with a cut off frequency of 20 Hz. All EMG data (including MVICs) were subject to a root mean squared (RMS) moving average of 25 ms in order to create a wave envelope. EMG data was normalised to maximum voluntary isometric contractions (MVIC); three MVICs were recorded for each muscle following completion of walking trials. The MVIC data was recorded against manual resistance for 5 s with a gradual build up of 2 s prior to maximal effort for the final 3 s. The peak value from a 500 ms window obtained from the 3-s maximal effort of the MVIC was used as the reference value similar to the methods used by Bogey et al. [17] and Murley et al. [18]. All participants were verbally encouraged in a standard manner during the MVICs and a 1-min recovery period was set between repetitions. Kinematic data were subject to a fourth order Butterworth low pass filter with a cut off of 6 Hz.

2.6. Statistical analyses

Statistical analyses were performed using SPSS 17.0 (SPSS Inc., Chicago, IL, USA). Demographic and group characteristics were summarised with the mean, and either standard deviation (SD) or range. Biomechanical and EMG data were normalised to 100% of stance and analysed using the absolute coefficient of multiple correlation (CMC) [19]. Within subject variation was assessed using two way mixed model intra class correlation coefficients (ICC) with 95% confidence intervals (CI) [20]. The descriptors poor, fair-to-good, and excellent reliability were assigned to ICC cut off values of <0.40, between 0.40 and 0.75, and >0.75 respectively [20], similar cut-offs were used for the CMC values. The standard error of measurement (SEM) with 95% CI was calculated to express the level of error in original and clinically meaningful units.

3. Results

3.1. Group characteristics

Five patients with RA (3F:2M) and PPV with a mean (SD) age of 53 (9) years and mean disease duration of 7 (5) years were recruited. Numerical differences were recorded between time points in global and foot related measures of disability (Table 1). Analyses were restricted to the stance phase of gait as preliminary analyses indicated that CMCs can be falsely inflated due to muscle inactivity during the swing phase.

3.2. EMG reliability

Within- and between-day reliability of muscle activation patterns are presented in Table 2. Within-day reliability was mostly excellent with only two muscles falling marginally below mean CMC values of 0.75; TP and peroneus longus. The lowest

Table 1
Summary of demographic, clinical and disease features.

Variable	RA week 0 ($n=5$)	RA week 1 ($n=5$)
Age – mean (range)	53 (41–65)	–
Gender (M:F)	2:3	–
Disease duration (years)	7 (3–15)	–
$FIS_{impairment\ subscale}$ (0–21)	11 (2–21)	11 (1–20)
$FIS_{disability\ subscale}$ (0–30)	11 (0–25)	12 (0–28)
HAQ	1 (0.4–2)	1 (0.2–2)
Foot pain VAS (0–100 mm)	32 (0–56)	26 (0–53)
General Health VAS (0–100 mm)	28 (0–52)	16 (3–36)
Arthritis VAS (0–100 mm)	27 (1–41)	21 (1–50)
Structural Index – rearfoot (0–7)	3 (2–4)	2 (2–3)
Structural Index – forefoot (0–12)	5 (1–8)	5 (1–8)
Swollen foot joint count (0–14)	0 (0–0)	0 (0–1)
Tender foot joint count (0–14)	1 (0–5)	3 (0–6)
Walking speed (m/s) BF/SH	1.18 (0.15)/ 1.22 (0.15)	1.21 (0.15)/ 1.21 (0.16)

M, male; F, female; FIS, Foot Impact Scale for RA; HAQ, Health assessment questionnaire; VAS, visual analogue scale; BF, barefoot; SH, shod.

Table 2
Summary of relative EMG reliability.

Muscle	Condition	CMC within-day (0)	CMC within-day (1)	CMC between-day
Medial gastrocnemius	Barefoot	0.756 (0.652–0.838)	0.825 (0.724–0.900)	0.764 (0.591–0.873)
	Shod	0.854 (0.768–0.926)	0.846 (0.811–0.887)	0.841 (0.777–0.891)
Peroneus longus	Barefoot	0.719 (0.580–0.852)	0.686 (0.486–0.857)	0.626 (0.396–0.776)
	Shod	0.809 (0.725–0.891)	0.687 (0.457–0.872)	0.670 (0.518–0.790)
Soleus	Barefoot	0.823 (0.732–0.881)	0.848 (0.813–0.922)	0.747 (0.579–0.827)
	Shod	0.850 (0.709–0.909)	0.861 (0.797–0.911)	0.769 (0.615–0.891)
Tibialis anterior	Barefoot	0.827 (0.726–0.947)	0.812 (0.765–0.859)	0.801 (0.706–0.879)
	Shod	0.859 (0.757–0.918)	0.862 (0.824–0.911)	0.837 (0.780–0.884)
Tibialis posterior	Barefoot	0.737 (0.623–0.848)	0.717 (0.656–0.838)	0.619 (0.502–0.684)
	Shod	0.771 (0.686–0.879)	0.797 (0.680–0.888)	0.697 (0.619–0.782)

CMC, coefficient of multiple correlation.

Table 3
Summary of EMG intraclass correlation coefficients and standard error of measurement.

Muscle/variable	Barefoot		Shod	
	ICC (95% CI)	SEM (95% CI)	ICC (95% CI)	SEM (95% CI)
Gastrocnemius peak MS/prop	0.10 (0, 0.83)[c]	13 (−20, 52)	0.47 (0, 0.92)[b]	16 (−32, 57)
Time peak MS/P	0.53 (0, 0.93)[b]	6 (−5, 30)	0.82 (0.18, 0.98)[a]	2 (−10, 3)
Peroneus longus peak contact	0 (0, 0.83)[c]	19 (−53, 51)	0 (0, 0.78)[c]	12 (−27, 40)
Time peak contact	0.91 (0.39, 0.99)[a]	1 (−4, 3)	0 (0, 0.04)[c]	5 (−16, 13)
Peak MS/P	0.19 (0, 0.87)[c]	17 (−36, 58)	0.03 (0, 0.81)[c]	18 (−29, 71)
Time peak MS/P	0.07 (0, 0.84)[c]	8 (−28, 15)	0 (0, 0.68)[c]	9 (−30, 19)
Soleus peak MS/P	0.51 (0, 0.94)[b]	15 (−41, 43)	0.51 (0, 0.94)[b]	18 (−48, 52)
Time peak MS/P	0.95 (0.63, 0.99)[a]	2 (−5, 4)	0.74 (0, 0.97)[b]	1 (−4, 4)
Tibialis anterior peak contact	0.55 (0, 0.94)[b]	4 (−9, 14)	0.09 (0, 0.85)[c]	5 (−10, 16)
Time peak contact	0.20 (0, 0.85)[c]	1 (−5, 2)	0.25 (0, 0.87)[c]	1 (−2, 5)
Tibialis posterior peak contact	0 (0, 0.74)[c]	27 (−60, 90)	0.53 (0, 0.94)[b]	11 (−33, 26)
Time peak contact	0.57 (0, 0.94)[b]	2 (−8, 3)	0.32 (0, 0.90)[c]	2 (−4, 7)
Peak MS/P	0 (0, 0.68)[c]	8 (−50,−6)	0.71 (0, 0.96)[b]	15 (−46, 35)
Time peak MS/P	0.45 (0, 0.92)[b]	4 (−8, 12)	0.87 (0.14, 0.98)[a]	1 (0, 5)

ICC, intraclass correlation coefficient; SEM, standard error of measurement; MS/P, combined midstance/propulsive phase.
[a] Excellent ICC.
[b] Moderate ICC.
[c] Poor ICC, SEM for magnitude variables expressed as % MVIC and SEM for temporal variables expressed as % stance.

mean value was 0.68 for peroneus longus. In all cases, both within- and between-days, the CMC values were greater in the shod condition than the barefoot condition. Between-day reliability ranged from moderate-to-good to excellent; values for peroneus longus and TP again fell below 0.75.

Discrete EMG variables in terms of timing and magnitude of contraction were investigated between-days (Table 3). Each muscle was analysed in the contact and combined midstance/propulsive phase [18]. A number of ICC values fell into the poor category (<0.40) and negative values were found in TP and peroneus longus. A negative ICC value is suggestive of greater levels of intra-session variation than inter session variation; where negative values occurred these are represented by 0. For the majority of studied variables the ICC results were poor, only four variables were in the excellent category and these were all in the temporal domain. The SEM varied across both the temporal and the amplitude characteristics in all muscle groups; however there was a trend towards reduced levels of error in the temporal domain.

Table 4
Within- and between-day relative reliability of kinematic data.

Variable	Condition	CMC within-day (0)	CMC within-day (1)	CMC between-day
Rearfoot Inv/Ev	Barefoot	0.856 (0.470–0.972)	0.943 (0.818–0.986)	0.676 (0.120–0.945)
	Shod	0.843 (0.628–0.978)	0.928 (0.863–0.982)	0.686 (0.343–0.972)
Rearfoot DF/PF	Barefoot	0.953 (0.890–0.990)	0.949 (0.789–0.993)	0.958 (0.911–0.980)
	Shod	0.987 (0.981–0.991)	0.981 (0.960–0.993)	0.957 (0.885–0.983)
Rearfoot Int/Ext Rot	Barefoot	0.848 (0.681–0.944)	0.858 (0.746–0.961)	0.548 (0.103–0.909)
	Shod	0.799 (0.560–0.978)	0.869 (0.745–0.948)	0.630 (0.374–0.938)
Navicular height	Barefoot	0.986 (0.968–0.998)	0.994 (0.988–0.998)	0.983 (0.958–0.992)
	Shod	0.993 (0.980–0.999)	0.996 (0.993–0.998)	0.992 (0.986–0.997)
Midfoot Inv/Ev	Barefoot	0.853 (0.562–0.978)	0.923 (0.832–0.970)	0.712 (0.315–0.935)
Midfoot DF/PF	Barefoot	0.932 (0.884–0.987)	0.964 (0.951–0.980)	0.893 (0.797–0.955)
Midfoot Abd/Add	Barefoot	0.820 (0.322–0.986)	0.926 (0.822–0.982)	0.557 (0.218–0.922)
Forefoot Inv/Ev	Barefoot	0.825 (0.698–0.985)	0.883 (0.823–0.955)	0.638 (0.120–0.965)
	Shod	0.854 (0.791–0.925)	0.846 (0.665–0.970)	0.563 (0.132–0.873)
Forefoot DF/PF	Barefoot	0.954 (0.914–0.994)	0.979 (0.962–0.990)	0.938 (0.900–0.975)
	Shod	0.956 (0.879–0.984)	0.963 (0.877–0.988)	0.897 (0.770–0.972)
Forefoot Abd/Add	Barefoot	0.881 (0.538–0.990)	0.953 (0.852–0.995)	0.746 (0.072–0.972)
	Shod	0.917 (0.817–0.988)	0.956 (0.906–0.994)	0.821 (0.542–0.971)

CMC, coefficient of multiple correlation; GRF, ground reaction force; Inv/Ev, inversion/eversion; DF/PF, dorsiflexion/plantarflexion; Int/Ext Rot, internal/external rotation; Abd/Add, abduction/adduction.

Table 5
Intraclass correlation coefficients and standard error of measurement for discrete kinematic variables.

Variable	Condition	ICC (95% CI)	SEM (95% CI)
Initial foot contact angle (deg)	Barefoot	0.81 (0, 0.98)	0.63 (0.01, 3.51)
	Shod	0.93 (0.53, 0.99)	0.73 (−0.78, 3.29)
Terminal stance PF angle (deg)	Barefoot	0.87 (0.34, 0.98)	1.78 (−2.35, 7.56)
	Shod	0.91 (0.36, 0.99)	0.89 (−0.74, 4.24)
Lowest navicular height (mm)	Barefoot	0.92 (0.45, 0.99)	1.10 (−3.56, 2.58)
	Shod	0.90 (0.47, 0.99)	1.11 (−4.35, 1.82)
Peak RF eversion angle (deg)	Barefoot	0.80 (0, 0.97)	1.99 (−5.05, 5.61)
	Shod	0.81 (0, 0.98)	1.93 (−5.05, 5.67)
Peak MF inversion angle (deg)	Barefoot	0.82 (0.04, 0.98)	1.68 (−5.57, 3.77)
Peak FF dorsiflexion angle (deg)	Barefoot	0.80 (0, 0.97)	0.60 (−1.81, 1.55)
	Shod	0.85 (0.19, 0.98)	0.86 (−2.95, 1.83)
Peak FF abduction angle (deg)	Barefoot	0.97 (0.78, 0.99)	0.76 (−1.58, 2.64)
	Shod	0.97 (0.76, 0.99)	0.85 (−1.97, 2.79)

ICC, intraclass correlation coefficient; SEM, standard error of measurement; deg, degrees; PF, plantarflexion; RF, rearfoot; MF, midfoot; FF, forefoot.

The SEM for magnitude of TP contraction ranged from 8% to 27% of MVIC across both phases of stance. For timing of peak TP activity the SEM ranged from 1% to 4% of the stance phase. Despite the smaller levels of error in the temporal domain, in almost all cases the 95% CI crossed zero across both domains.

3.3. Kinematic reliability

Within-day reliability was excellent with all kinematic variables reaching mean CMC values >0.75 in both conditions (Table 4). There was a general trend towards improved reliability with higher CMC values in the shod condition compared to barefoot for the majority of variables. Between-day reliability ranged from fair-to-good to excellent for all kinematic variables with the lowest value of 0.548 recorded for the rearfoot in the transverse plane. There was a trend towards higher CMC values in the sagittal plane compared to frontal and transverse planes within- and between-days.

Between-day kinematic ICCs and SEM are presented in Table 5. Values were excellent ranging from 0.89 to 0.99 for all variables with low SEM values however, in the majority of cases the 95% CI of the SEM crossed zero. There was no discernible pattern or differences noted between barefoot and shod conditions.

4. Discussion

This study set out to assess the within- and between-day reliability of EMG activity patterns for TP and other selected lower limb muscles as well as 3D multi-segmented foot kinematics in an RA cohort with PPV. The EMG and kinematic results demonstrated superior within-day reliability to between-day reliability in this small cohort. Moreover, there was a trend towards improved reliability in the shod condition. However, in this patient cohort discrete variables for muscle timing and amplitude demonstrated unacceptably poor levels of reliability between-days.

Tibialis posterior is the most powerful supinator of the rearfoot, it also adducts the forefoot and contributes to ankle plantarflexion; in addition it acts as a dynamic stabiliser of the medial longitudinal arch. Increased EMG signal amplitude and activity during the stance phase of gait has been consistently demonstrated in RA and non-RA PPV cohorts [7–9] this suggests the muscle is working to counteract the motion tendencies of eversion and dorsiflexion in the ankle/subtalar complex and forefoot abduction which characterise PPV. We recorded TP EMG patterns in RA patients with PPV and demonstrated that this could be reliably measured using a fine-wire, indwelling electrode approach in a single session both barefoot and in-shoe. However, while the overall EMG muscle pattern was similar 1 week later, discrete timing and magnitude variables were not reliable. Altered muscle activity has also been described for other lower leg muscles including gastrocnemius, peroneus longus, peroneus brevis and flexors digitorum and hallucis longus [7–9]. This evidence suggests these muscles must be studied if the complexity of the primary impairment and compensatory mechanisms are to be fully understood. However, we observed the same trend for poor reliability of other lower limb muscles as for TP: this confirms and extends the observations of Murley et al. [21] to include an inflammatory joint disease.

Factors related to EMG technique, the effects of RA on EMG measurement, and the effects of RA directly on muscle structure and function should be considered when interpreting these results. Potential sources of error with EMG are well described and include subcutaneous soft-tissue volume, cross talk, motion artefact, intramuscular bleeding [22], accuracy of electrode placement, skin preparation [15] and retraction of electrodes during dynamic tasks [23]. In an attempt to minimise their effect we employed a rigorous protocol adapting best practice, including the SENIAM guidelines. However, there are specific issues related to RA which may compound these potential sources of error. RA is associated with metabolic changes leading to loss of muscle mass and strength [24]. Moreover, muscle activation capacity can be reduced depending on disease activity state and joint pain and effusion [25,26]. Structural muscle changes such as decreased volume may have influenced the precision of surface and fine-wire electrode placement and signal detection. Effusions or pain in the ankle, subtalar or midtarsal joints where the lower leg muscle tendons cross to insert into the foot may have influenced muscle activation patterns. Changes in RA disease activity state with up or down regulation of inflammatory cytokines may have altered muscle physiology and EMG signal detection. We did not measure muscle cross-sectional areas or volume so cannot account for the influence of structural changes. However, we attempted to limit error by guiding the electrode to the TP muscle belly using ultrasound; a technique we have shown to be highly accurate [27].

RA is a disease with a variable course, characterised by flares and remissions with associated fluctuations in patient symptoms. We found evidence that self-reported disease state and pain changed between testing periods so this may have changed muscle physiology, muscle activity, and subsequently EMG signal detection. Muscle-specific force and muscle activation patterns have been found to be normal in RA patients with stable disease, even with significant muscle loss [28]. In our cohort disease activity may have fluctuated but a firmer conclusion can only be reached on this if objective measures were employed. Finally, the RA patients in this cohort did report joint tenderness but no swelling. We detected difference in joint tenderness between time points and this may have adversely influenced reproducibility of muscle activity capacity during MVIC tests, as well as muscle function during gait. Adapted gait patterns in RA are well described [10,29].

RA patients seldom walk barefoot and this may explain the superior reliability in the shod condition.

Combining EMG muscle activation patterns with multi-segmented foot kinematics presents an opportunity to understand the relationship between muscle function and joint motion. The reliability of the kinematic data in this study was consistent with that reported for other inflammatory joint conditions [4,13]. We confirmed previous observations where larger foot segments have greater reliability both within- and between-days in the sagittal plane in comparison with smaller segments with smaller ranges of motion, particularly in the frontal and transverse planes. Inflammatory joint diseases present challenges for identification of surface landmarks, in particular when joints are swollen, tender and deformed [13]. Nevertheless, the model presented here showed excellent reliability for discrete variables both barefoot and shod.

There are a number of limitations in this study. Firstly the complexity and time burden of the protocol restricted the sample size, particularly in a heterogeneous disease such as RA. Future work should attempt to recruit a larger sample in order to draw more meaningful conclusions. Secondly, normalising the EMG signals to MVICs has inherent limitations in a patient population with fluctuating symptoms, particularly joint pain and effusions. Alternative techniques such as submaximal contractions of dynamic contractions merit future study [21]. Thirdly, while the use of the intensity and duration of muscle activation are appealing variables to study for clinical and research applications, other forms of analysis may show greater reliability.

In conclusion, patterns of muscle activation and kinematic motion appeared more consistent than discrete variables and absolute measures of error. The findings demonstrate that, in this cohort of RA patients using these EMG variables, the use of discrete EMG variables between time points is not supported either barefoot or shod. Within session reliability is greater than between session and this should be considered when planning intervention or longitudinal studies; a single session evaluation or alternative analysis may be more appropriate due to the lack of measurement precision. Kinematic reliability has been established in the presence of pathology in this cohort but should be interpreted within the error limits.

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Conflict of interest

The authors have no conflict of interest to declare.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.gaitpost.2012.05.008.

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