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Changes in 3D Joint Kinematics Support the Continuous Use of Orthoses in the Management of Painful Rearfoot Deformity in Rheumatoid Arthritis

JAMES WOODBURN, PHILIP S. HELLIWELL, and SHARON BARKER

ABSTRACT. ***Objective.*** To evaluate the efficacy of custom foot orthoses for the management of painful rearfoot valgus in patients with rheumatoid arthritis (RA).

Methods. Patients were randomized to receive custom-manufactured rigid carbon graphite foot orthoses (RA-orthosis) or enter a control group (RA-control) receiving no orthotic intervention. Three-dimensional (3D) kinematics were measured at the ankle joint complex (AJC) using an electromagnetic tracking (EMT) system under barefoot, shod, and orthosis walking conditions. Previously established normal 3D kinematic data were used to descriptively compare motion patterns in both RA groups and statistical analyses were performed on integrals of motion-time for each axis of rotation from data collected at baseline, 3, 6, 12, 18, 24, and 30 months.

Results. Compared with healthy control subjects, all patients with RA demonstrated excessive subtalar joint eversion motion through the stance phase of gait ($p < 0.0001$) coupled with excessive internal leg rotation ($p < 0.0001$). Custom-manufactured orthoses significantly reduced eversion through stance ($p = 0.009$) and re-established equilibrium of motion relative to neutral joint position. Correcting the frontal plane component of the deformity did not lead to a significant reduction in internal leg rotation ($p = 0.294$). The devices had no effect on tibiotalar dorsiflexion/plantarflexion ($p = 0.960$). Prospectively, the rigid orthoses maintained and then improved the reduction in cumulative subtalar eversion motion ($p < 0.0001$). Minimal changes in cumulative subtalar component eversion and internal leg rotation were recorded for both RA groups when walking barefoot but the effect was significantly less for the RA-control group. From 12 months onwards, internal leg rotation started to decrease, suggesting re-coupling of motion, but the overall motion pattern remained abnormal in comparison with normal reference values.

Conclusion. These results support the continuous use of custom-manufactured foot orthoses to correct deformity and optimize AJC function in RA patients with early painful deformity of the rearfoot. (J Rheumatol 2003;30:2356–64)

Key Indexing Terms:
FOOT ORTHOSES — THREE-DIMENSIONAL KINEMATICS — RHEUMATOID ARTHRITIS — RANDOMIZED CONTROLLED TRIAL

The ankle joint complex (AJC) (tibiotalar and subtalar joints) provides a linkage between the leg and foot, and these joints must be stable yet flexible to permit normal locomotion. Under normal circumstances the articular surface geometry, the surrounding ligaments, joint capsules and other soft tissues, and the surrounding musculature both guide and limit AJC motion[1,2]. In rheumatoid arthritis (RA), synovitis, effusion, and erosive arthropathy of the tibiotalar, subtalar, and talonavicular joints as well as tibialis posterior tenosynovitis are thought to combine to cause clinically recognizable valgus heel or pes planovalgus deformity[3-5]. The progression rates for deterioration in AJC function in RA have not been properly studied. Some believe the deformity can arise in early RA where the pathogenesis is thought to involve only ligament, joint capsule, and other soft-tissue deformation in response to normal external loads in the presence of localized synovitis and effusion[6,7]. It is during this period, often missed in routine clinical practice, that calls have been made to intervene with appropriate foot orthotic treatment[7,8].

Evidence for localized tibiotalar, subtalar, and midtarsal joint synovitis and erosion and tenosynovitis of tibialis

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posterior has been found[4,5,9,10]. There is also strong evidence that rearfoot deformity in RA is associated with significant localized foot pain and impairment of locomotion[3,11]. There is, however, neither evidence for, nor any technique described, to accurately quantify localized soft-tissue laxity. Pragmatically, patients are selected for early orthotic management on the basis of clinical and/or imaging evidence of peritalar disease and valgus/pes planovalgus deformity that is flexible and where the joints show normal range of motion. Temporary reduction in foot pain and increased gait function in response to foot orthoses has been shown in several uncontrolled studies[12-15]. However, one randomized controlled clinical trial found no significant clinical benefits for foot orthoses over placebo in a group of men with advanced RA[16]. In contrast, we have shown significant clinical improvements in foot pain and disability when foot orthoses were used in patients with early painful rearfoot deformity, with the effect sustainable for 30 months[17].

Despite evidence of clinical response, there have been no published studies, as far as we are aware, investigating the changes in rearfoot joint motion in RA as part of orthotic treatment. Evidence exists that normal rearfoot motion can be restored using foot orthoses and modified footwear in other clinical conditions, chiefly the over-pronating runner, although the effect is variable[18–20]. Testing the hypothesis that custom foot orthoses change AJC 3-dimensional (3D) kinematics, motion analysis was conducted over 30 months in 2 RA groups randomized to receive either nothing or custom foot orthotic intervention.

MATERIALS AND METHODS

Study design. The study was a prospective, randomized, controlled clinical trial. At baseline patients with RA who had localized rearfoot pain and deformity were identified and randomized to an orthotic intervention (RA-orthosis) or control (RA-control) group. 3D kinematics were measured at the AJC at baseline and repeated intervals for a duration of 30 months. Reference 3D kinematic data were collected from a sex and age matched cohort of healthy adults (n = 45) measured previously in our gait laboratory[7]. Local research ethical committee approval was granted for this study. A complete description of the trial design, interventions and trial profile has been published[20].

Patients. Ninety-eight patients with RA (satisfying the 1987 American Rheumatism Association revised criteria for RA) were recruited from the Rheumatology Outpatient Clinic at St. Luke's Hospital, Bradford, UK[21]. Inclusion criteria were bilateral arthritis of the peritalar joint complex and valgus deformity of the heel, correctable on passive range of motion testing. Patients were excluded if they had concomitant musculoskeletal disease; central or peripheral nervous system disease and endocrine disorders, especially diabetes mellitus; rigid or semirigid rearfoot deformity; previous orthopedic foot surgery or foot orthotic treatment, or inappropriate footwear at time of recruitment.

Interventions. Our understanding of rearfoot function in RA was previously determined in kinematic studies and we used this knowledge to design a standardized orthosis. The orthoses were constructed on a neutral cast of each foot in rigid carbon-graphite (Super-Lyte®, Langers Biomechanics Group, Cheadle, UK) and included a deep heel cup and a contoured medial longitudinal arch to support the heel, subtalar, and midtarsal joints. Frontal plane correction was achieved by medial intrinsic forefoot and rearfoot posting, individually specified for each patient. Physicians, blinded to inclusion of patients in the study, were allowed to prescribe any type of foot orthosis for patients in the control group during routine outpatient followup.

Measurement of ankle joint complex 3D kinematics. An electromagnetic tracking system (EMT) (6DRESEARCH Skill Technologies Inc, Phoenix, Arizona, USA) was used to measure 3D joint motion. This system employs Fastrak® sensors (Polhemus Inc, Colchester, Vermont, USA) integrated with custom-designed kinematic software. The system tracks the position and orientation of the sensors with 6 degrees-of-freedom. Sensors were attached on the skin overlying the medial tibial surface between the midline of the knee and ankle joints and the posterior surface of the calcaneus[7,22]. A "bore-sighting" or neutral orientation alignment procedure was undertaken to rotate and align the axis reference frames for the transmitter source and sensors. This was conducted with the patient standing upright, with the foot and leg forming a 90º angle and the foot wedged on the plantar aspect to hold, following manual palpation and positioning, the subtalar joint in its neutral position[22]. After an initial familiarization period, patients were requested to walk at normal speed over an 8 m distance, passing through a transmitter-generated low-strength electromagnetic field. Computer software was used to detect the position and orientation of the EMT sensors through the electromagnetic field. After filtering raw data with a 6Hz low-pass digital Butterworth filter, further software routines calculated joint coordinate system angles as defined for the AJC by Siegler, *et al*[23].

3D motion was simultaneously recorded for the left and right AJC for 5 trials under 3 conditions for the RA-orthosis group (RA-orthosis group barefoot; RA-orthosis group shod; RA-orthosis group with orthosis) and 2 conditions for the RA-control group (RA-control group barefoot; RA-control group shod). To accommodate the heel mounted sensor inside the shoe, a window was created in the heel area of standard stock shoes, which all patients wore[7,22]. Synchronous gait cycle timings were identified using heel and forefoot skin-mounted pressure switches (Figure 1).

Prior to data collection all metal objects within 5 m of the gait equipment were removed since electromagnetic tracking is highly sensitive to metallic interference. Within this carefully prepared measurement volume the system had an accuracy of 0.75° for angular rotation. Careful choice of anatomical sites for sensor mounting resulted in negligible skin movement artefact as determined in a previous study[24]. Within and between-day coefficients of multiple correlation of between 0.97 to 0.77 were found for all 3 axes of rotation in both normal and RA patients, indicating good to high levels of precision[22,24].

Statistical analysis. 6DNORM software (M. Cornwall, Northern Arizona University, USA) was used to generate motion-time curves, normalized to 100% of the gait cycle, for each axis of rotation (dorsiflexion/plantarflexion in the sagittal plane, inversion/eversion in the frontal plane, and internal/external rotation in the transverse plane). 3D motion under barefoot, shod, and orthosis conditions in the RA groups were descriptively compared to normal reference data[7]. The time a joint spends in an abnormal position during stance may be an important metric in comparison with a peak joint excursion, which may be a transient event. To address this we calculated the total joint excursion from the zero neutral position for a normalized time period, in degrees, as a single variable for analysis. This was done by summing, using the trapezium rule, positive and negative areas under the curve (Figure 2). Analysis of variance (ANOVA) with *post-hoc* tests was used to compare the baseline and followup motion-time integrals between the RA-orthosis and RA-control groups using factorial and repeat measure designs. All patients were analyzed according to group assignment, and missing data were replaced by last value carried forward. The significance level used was $p < 0.05$, and all analyses were conducted using SPSS 9.0 for Windows (SPSS, Chicago, Il, USA).

RESULTS

Fifty patients were prescribed custom orthoses and 48 patients were allocated to the control group (Table 1). At 30

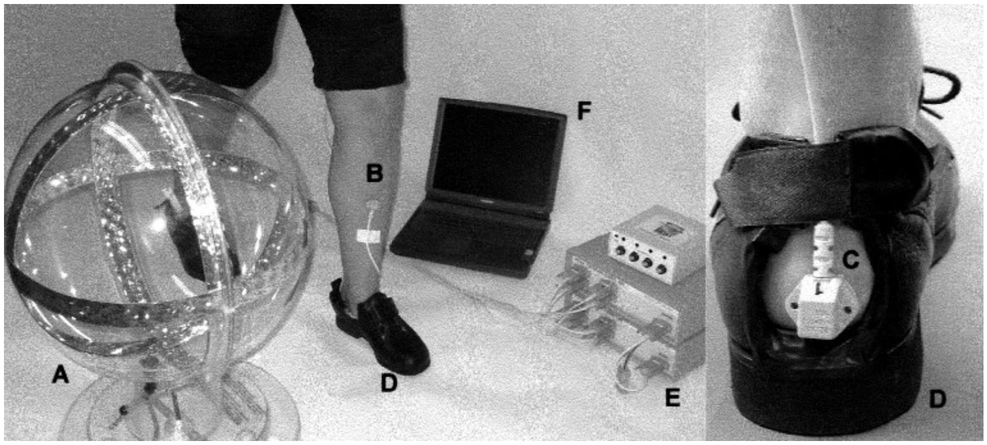

Figure 1. Arrangement diagram of the electromagnetic tracking system. A: transmitter; B: tibial sensor (hidden); C: calcaneus sensor; D: adapted stock shoe; E: motion capture unit; F: computer workstation.

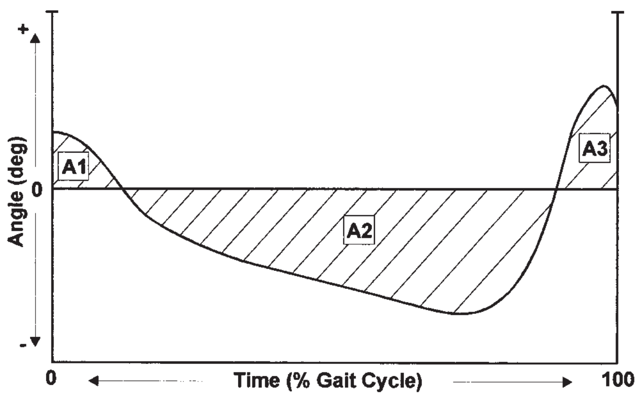

Figure 2. An example of a motion-time curve for inversion-eversion motion in the frontal plane for the ankle joint complex. The motion-time integral is calculated by summing the areas under the curve using the trapezium rule. In this example there are 2 phases where the joint is inverted, A1 and A3 with positive area units and 1 phase where the joint is everted, A2 with negative area units. A1 (+), A2 (–), and A3 (+) are summed to give the motion-time integral.

months, 43/50 (86%) in the orthosis group and 38/48 (79%) of patients in the control group completed the trial. Three patients in the control group were prescribed orthoses from hospital appliances or podiatry over the duration of the study. Normative reference 3D kinematic data were gathered from 45 age and sex matched healthy adults (mean age 51.8 years ± 12.4 years, 16 men and 29 women).

Baseline dorsiflexion/plantar flexion. At baseline, dorsiflexion/plantarflexion motion under barefoot conditions was characterized by 3 phases of motion during stance: an initial plantarflexion from heel strike to foot flat, dorsiflexion through mid-stance, and heel lift and rapid plantarflexion towards propulsion (Figure 3). The motion patterns for both RA groups were similar in shape to the normal group. There was no statistically significant difference in the mean motion-time integral between the RA-control (–61.0° ± 404.6°), RA-orthosis (69.5° ± 405.6°), and normal (–36.2° ± 326.7°) groups, p = 0.167, Figure 4. Under shod conditions, the motion patterns remained similar but were shifted in a plantarflexion direction with large negative motion time integrals reported in all groups (normal: –573.0° ± 307.0°; RA-control: –588.2° ± 423.9°; RA-orthosis: –523.3° ± 393.6°), Figure 4. *Post-hoc* analyses revealed significant differences between barefoot and shod dorsiflexion/plantarflexion motion for normal and both RA groups (p < 0.0001 all tests).

Baseline inversion/eversion. Inversion/eversion motion in the normal group was characterized by an inverted heel strike position, followed by eversion motion through the neutral joint position past mid-stance and early heel lift (Figure 3). During late stance, rapid inversion motion brought the joint past the neutral position into an inverted position during propulsion. The overall shape of the inversion/eversion motion pattern was similar for both RA groups, but shifted negatively (into eversion) on the ordinate in comparison to normal motion. As a consequence, the normal group had a small positive mean motion-time integral (87.3° ± 244.3°) in comparison with large negative integrals for the RA-control (–546.1° ± 282.6°) and RA-orthosis (–563.4° ± 366.7°) groups, and this was statistically significant (p < 0.0001), Figure 4. *Post-hoc* analysis showed significant pairwise differences between normal and RA-control (p < 0.0001) and normal and RA-orthosis (p < 0.0001), but not between the RA groups (p = 0.757). Shod motion differed significantly for all 3 groups (p < 0.0001 for all comparisons), characterized in a positive shift in the motion time integral: normal (247.8° ± 330.7°), RA-control (–252.3° ± 331.2°), and RA-orthosis (–264.1° ± 387.7°).

Baseline internal/external rotation. Normal AJC internal/external rotation showed internal rotation from a neutral

Table 1. Demography and baseline disease characteristics of the study participants.

Characteristic	RA Orthosis, n = 50	RA Control, n = 48
Age (yrs), mean (SD)	54.0 (11.8)	53.1 (11.1)
Male/female	16/34	17/31
Caucasian, n	45	45
Body mass, mean (SD)	73.7 (15.1)	73.2 (12.6)
Disease duration (yrs), median (range)	3 (1, 7)	3 (2, 6)
Foot function index (0-100 mm VAS), mean (SD)	41.1 (20.3)	33.6 (21.5)
Disease activity score, mean (SD)	3.4 (1.2)	3.2 (1.6)
HAQ (0–3), median (range), n	1.00 (0.47, 1.75)	1.00 (0.38, 1.75)

VAS: visual analog scale; HAQ: health assessment questionnaire.

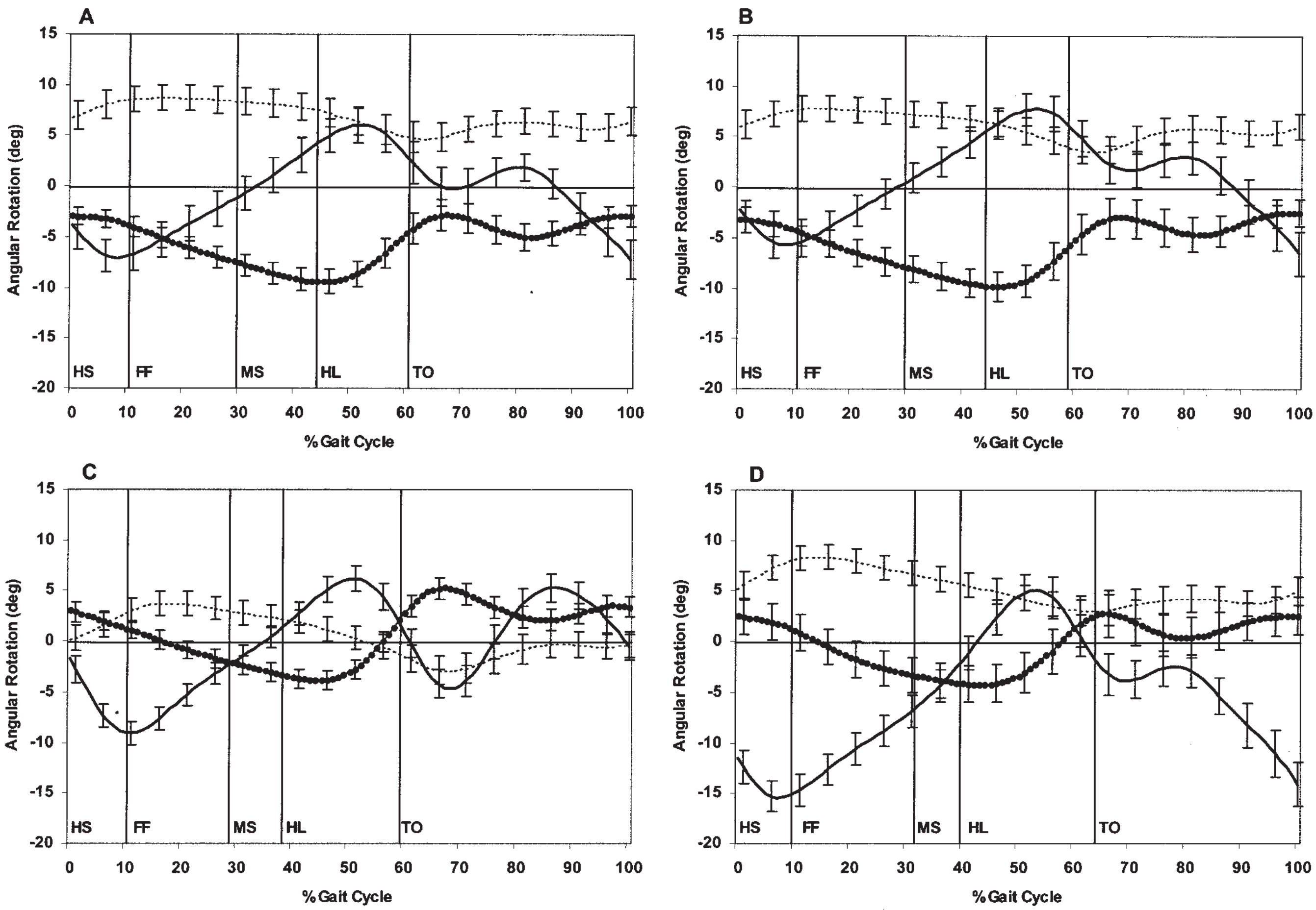

Figure 3. Motion curves for (A) RA-control group walking barefoot, (B) RA-orthotic group walking barefoot, (C) normative data for walking barefoot, (D) RA-orthosis group walking with orthoses. The solid line represents dorsiflexion(+) / plantarflexion(–), the solid line with markers represents inversion(+) / eversion(–), and the broken line represents internal(+) / external rotation(–). Gait events include HS: heel strike, FF: foot flat, MS: mid-stance, HL: heel lift, and TO: toe off. Bars represent the 95% confidence interval of the mean.

heel-strike position during the loading response, and slow external rotation through mid-stance and heel-lift. At propulsion, the joint was externally rotated past the neutral alignment. In both RA groups abnormal motion was captured: the AJC internally rotated from an excessive internally rotated heel strike position. External leg rotation was recorded through stance phase and propulsion but was insufficient to allow the joint to reach its neutral configuration (Figure 3). The motion-time integrals were significantly different between the normal (65.7° ± 335.4°) and both the RA-control (695.0° ± 357.7°, $p < 0.0001$) and RA-orthosis (610.1° ± 361.0°), but not between the RA groups ($p = 0.757$), Figure 4. Shod gait significantly reduced the amount of internal joint rotation for the normal group (12.6° ±

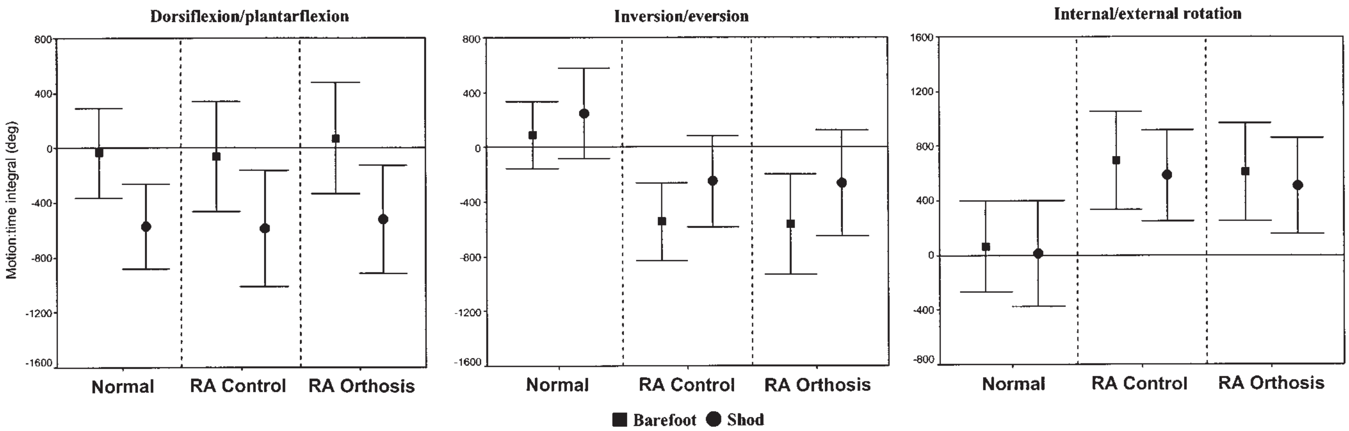

Figure 4. Mean 3-D kinematic motion:time integrals for barefoot and shod gait walking for normal, RA-control and RA-orthosis groups at baseline. Bars represent one standard deviation above and below the mean.

390.0°, p = 0.019), the RA-control group (584.2° ± 336.1°, p < 0.0001), and the RA-orthosis group (p < 0.0001).

Baseline effect of orthoses. The immediate effect of the foot orthoses on the motion-time integrals is shown in Figure 3D. The dorsiflexion/plantarflexion motion retained 3 phases, similar to the normal motion pattern, showed an increased range of motion over barefoot and shod conditions, but was shifted negatively on the ordinate. The orthoses restored normal phasic periods of eversion and inversion motion relative to the neutral joint position during the stance phase. At heel-strike and terminal propulsion the AJC was inverted and peak eversion towards heel lift was also reduced. The dorsiflexion/plantarflexion motion-time integrals were significantly different between barefoot (69.5° ± 405.4°) and shod (–523.3° ± 393.6°) p < 0.0001, and barefoot and orthosis conditions (–544.6° ± 372.7°) p < 0.0001, but not between shod and orthosis conditions (p = 0.960) in the RA-orthosis group (Figure 5). The inversion/eversion motion-time integral was significantly different between barefoot (–563.4° ± 366.7°) and shod (–264.1° ± 387.7°) p = 0.001, barefoot and orthosis (–26.1° ± 454.7°) p < 0.0001, and shod and orthosis conditions (p = 0.009). The orthoses had no significant effects on internal/external rotation (p = 0.294). In comparison with shod data from normal (p = 0.001) and RA-control groups (p = 0.009), only inversion/eversion motion was significantly different with orthotic intervention.

Longitudinal changes. The mean motion-time integrals for the 30-month duration of the study are presented in Figure 6. The observed baseline differences in dorsiflexion/plantarflexion motion between the barefoot, shod, and orthotic conditions were maintained for the duration of the study (p = 0.005). Reported baseline differences in inversion/eversion motion between the RA-control and RA-orthosis groups were maintained for the duration of the study (p < 0.0001) and between barefoot, shod, and orthotic conditions (p < 0.0001). A significant time effect was shown (p < 0.0001) whereby a positive shift in the motion-time integral

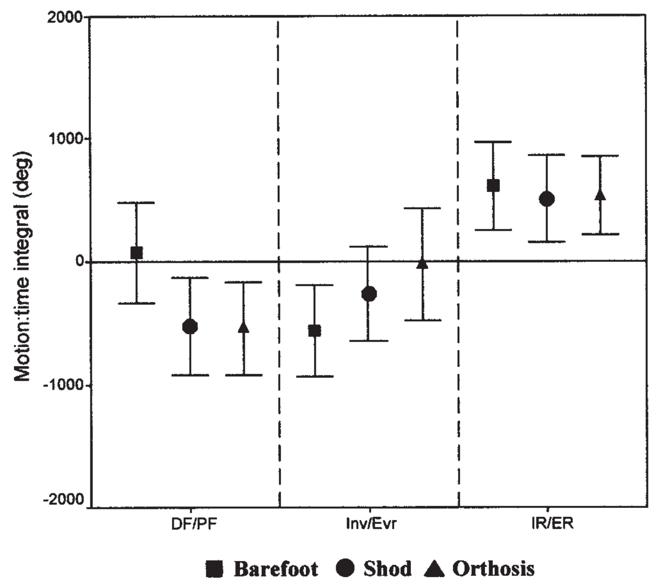

Figure 5. Mean 3-D kinematic motion:time integrals for barefoot, shod, and orthosis walking conditions in the RA-orthosis group at baseline. DF/PF: dorsiflexion/plantarflexion; Inv/Evr: inversion/eversion; IR/ER: internal/external rotation. Bars represent one standard deviation above and below the mean.

was seen, especially between baseline and 12 months (p < 0.0001) and then between 12 and 30 months (p < 0.0001). This trend differed between RA groups (p = 0.030), in both barefoot and shod conditions. A sharp increase in the motion-time integral was found between 12 and 30 months in the RA-orthosis group, whereas the RA-control group showed only a slight or no increase in the motion-time integral during this period. Significant differences between shod and barefoot internal/external AJC rotation found at baseline were maintained for the duration of the study (p = 0.006). Over 30 months, the RA-orthosis group showed a significant reduction in internal rotation in comparison with RA-control (p = 0.007). A significant time effect was found (p <

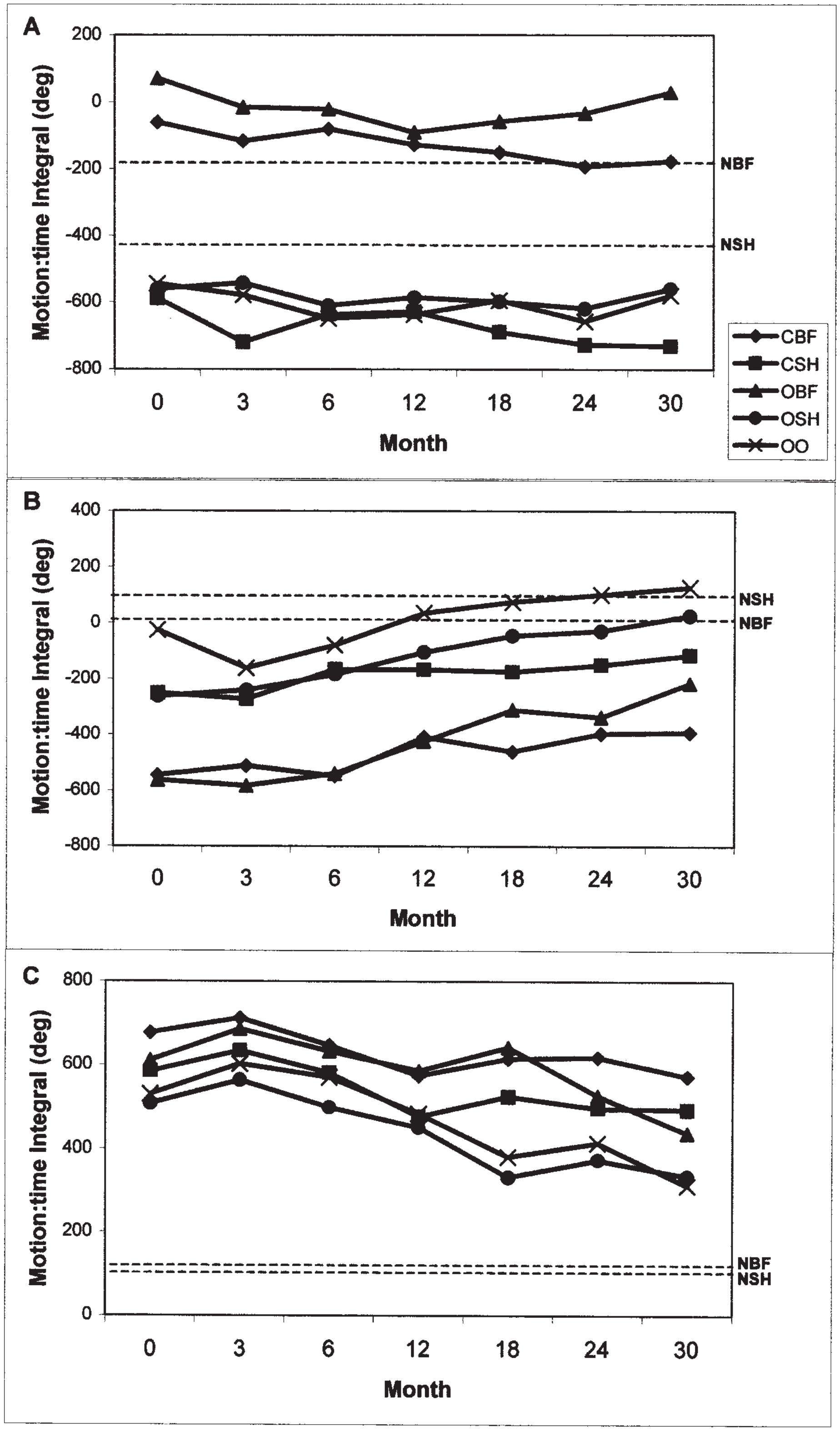

Figure 6. Mean 3-D kinematic motion:time integrals for A: dorsiflexion/plantarflexion; B: inversion/eversion; and C: internal/external joint rotations from baseline to 30 months. CBF: RA control group barefoot; CSH: RA control group shod; OBF: RA-orthosis group barefoot; OSH: RA-orthosis group shod; OO: RA-orthosis group with orthosis; NBF: normal mean reference motion:time integral for barefoot walking condition; NSH: normal mean reference motion:time integral for shod walking condition.

0.0001) characterized by a reduction in the motion-time integrals between baseline and all intervals from 6 to 30 months, the largest changes occurring for the RA-intervention group under barefoot, shod, and orthotic walking conditions.

DISCUSSION

We tested the hypothesis that custom foot orthoses change AJC kinematics in patients with RA. Using EMT, normal AJC motion was determined for healthy adults and the patterns, magnitudes, timings, and repeatability were in agreement with other published literature[25,26]. Furthermore, prior to orthotic intervention, both RA groups had abnormal AJC motion characterized by excessive eversion and loss of inversion about the neutral joint position. Our data using 3D motion analysis support and add to the reports of Keenan, *et al* and other investigators who used uniplanar motion analysis[3] and single foot models with individual cases[27] to show similar changes to foot motion in RA. Following the introduction of the custom foot orthoses, the cumulative amount of eversion motion was reduced through the stance phase. Moreover, the orthoses re-established equilibrium of motion through the neutral subtalar joint position. An inverted heel strike position was re-established changing the initial AJC inclination by a mean of 5.7°. Peak eversion at mid-stance was reduced on average by 4.6° and inversion motion during terminal stance, i.e., passing through neutral subtalar joint position to an inverted position, represented a mean change of 8.5°. Importantly, the cumulative frontal plane changes with orthoses, as measured by the integral of motion, were significantly greater with shoes and orthoses than with shoes alone. In accordance with the findings from other studies, the orthotic effect was at its greatest during loading response and terminal stance periods[28]. Here the AJC is most vulnerable to becoming unstable because the articular surfaces are not fully loaded, so the intervention effect is desirable[29].

Improvement in 3D kinematics at the AJC was accompanied by significant and sustainable reduction in foot pain and disability, as reported in the clinical component of this study[20]. What mechanisms related to the changes in 3D joint kinematics are associated with these beneficial clinical effects? Motion control is only one aspect of joint stabilization, and our data cannot directly address issues of whether these kinematic changes altered the motion guiding and stability properties of ligaments, joint capsule, and fascia surrounding the AJC, or reduced focal articular and soft-tissue stresses, or whether proprioception and neuromuscular control around the AJC were improved. A recent study using 3D magnetic resonance imaging foot reconstructions showed subtle differences in the architecture of the rear and midfoot between RA patients with peritalar disease and rearfoot deformity[30]. Combined with the kinematic data, this evidence supports our clinical impression that custom orthoses are preferable because they are manufactured on plaster models that capture more precisely the individual foot morphology to optimize orthotic fit and hence functional control. While mutually dependent motion in the rear and forefoot were controlled by the deep heel cup and medial posting, motion control is also provided by the rigidity of the carbon-graphite and its close conformity to the corrected shape of the medial longitudinal arch. Many patients with rearfoot disease have accompanying disease in the midfoot, especially around the talonavicular joint, and the orthoses may have resisted medial longitudinal arch collapse. Presently, the kinematic measurement system does not permit access to the midfoot or forefoot with in-shoe measurement, and we are unable to assess these important functional aspects. In further support of custom-manufactured orthoses, *in vitro* modeling of pes planovalgus deformity found < 2% improvement in arch height with limited correction of rearfoot valgus for over-the-counter devices[31].

While we are encouraged by these initial findings, the foot orthoses provided functional control for only one plane of motion. At baseline there was no evidence that the orthoses reduced excessive internal leg rotation. This suggests that correction of frontal plane rotation does not induce coupled motion, that is, increased subtalar inversion did not couple with external leg rotation. The coupling action is determined by a number of factors including joint congruency, ligament tautness, and neuromuscular control. Our results suggest that in RA soft-tissue laxity, incongruent joint surfaces, and perhaps also tibialis posterior dysfunction render failure in the coupling mechanism during the initial phase of foot orthotic management. This merits further investigation and also suggests that physical rehabilitation targeting these mechanisms should be included in an early intervention program.

To the best of our knowledge, the longterm functional properties of foot orthoses have not been reported. Our findings show that continuous orthotic therapy may afford longterm stability to the AJC. Excessive eversion was reduced over a period of 30 months under both shod and orthotic conditions in the intervention group. Most interestingly, barefoot eversion was also significantly reduced between 12 and 30 months. We speculate that during the initial 12 months, orthotics improved the orientation and alignment of soft-tissue structures in and around the AJC and improved proprioception and neuromuscular control while maintaining joint flexibility. Furthermore, from 12 to 30 months, internal leg rotation was sharply reduced in the orthotic intervention group, especially under barefoot walking conditions. This suggests that some degree of recoupling occurred that required either a threshold of frontal plane correction to be reached to induce coupling or improved soft-tissue function or a combination of both factors.

An entry criterion for all RA patients was good footwear to accommodate an orthosis. Although not formally recorded, our impression was that most patients in the control group wore fully enclosed footwear with medial heel counter support. Partial control of excessive eversion was found at baseline with shoes in the RA control group. This group had median disease duration of 3 years and we recorded increased frequency of disease modifying antirheumatic drugs (DMARD) and combined DMARD therapy over the course of 30 months. Subsequently, and to our surprise, the control group showed no evidence of progressive AJC deformity. In fact, slight improvements in cumulative subtalar component eversion and internal leg rotation were recorded, but the effect was significantly less than for the orthotic intervention group. The combined management of underlying inflammation and the basic mechanical action of ordinary footwear may have served to prevent worsening of the condition. Furthermore, no confounding from adjunct therapy was found and only 3/48 patients received any kind of insole/orthotic treatment from routine clinical followup over the course of the study.

We acknowledge that the joint excursions were small, but good system precision and accuracy permitted adequate detection of treatment effect. Skin movement artefact may contribute to measurement error and was reduced to systematic error in this study since the sensors were not removed or repositioned between test conditions following boresight.

Kinematic studies have previously evaluated the effect of foot orthoses for manipulating rearfoot pronation in otherwise healthy individuals and in intervention studies in over-pronators, both predominantly under running conditions[17-19,32-35]. These studies are useful because they have shown varied response in terms of magnitude and planar dominance for main orthotic effect, as seen here, but direct comparison is difficult. In RA it would appear that foot orthoses have an important role in correcting deformity at the tibiotalar and subtalar joints. Since deformity varies in severity and complexity depending on the number of rear and midfoot joints involved we prefer to use custom-manufactured devices. Maintaining joint stability may also reduce internal joint and surrounding soft-tissue stresses. What remains unclear is the effect of changing joint function on localized inflamed synovium. Future studies will be directed towards quantifying localized rearfoot synovitis and evaluating response to orthotic and other localized interventions. Comparisons are also required, including health economics as an outcome, of custom-manufactured, pre-manufactured, simple insoles, and CAD-CAM orthoses.

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