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The effect of knee prosthesis design on tibiofemoral biomechanics during extension tasks following total knee arthroplasty

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ABSTRACT

Background: Determine whether the tibiofemoral motion and electromyographic activity of the knee differs in patients with a medial pivot implant, compared to those with cruciate-retaining and posterior-stabilised designs, during knee extension after Total Knee Arthroplasty (TKA).

Methods: An observational study was conducted on a cohort of patients that had undergone TKA for a minimum of 12 months prior. Three matched groups ($n = 18$) were categorised based on implant type: medial-pivot (MP), posterior-stabilised (PS) and cruciate-retaining (CR). Kinematics, with motion analysis (Vicon, USA) and surface electromyography (Delsys, USA) were assessed during step-ascent and walking tasks.

Results: All groups displayed a similar amount of knee extension in both tasks. They also paradoxically produced an average mean internal rotation movement during knee extension in both the step-ascent and walking tasks. The only significant difference was found in the step-ascent task, in which the MP group produced a larger absolute amount of rotation than the CR implant group ($P = 0.007$), but neither group differed from the PS implant group. The groups did not differ in rotation during the walking task ($P > 0.05$). The MP group displayed significantly ($P < 0.01$) greater knee extensor activation during the step-ascent than the PS group.

Conclusion: The MP design was only significantly different to another implant design for the step-ascent task. Patients with either knee implant types were not strictly limited to producing the traditional “screw-home” mechanism, defined by external rotation during extension. Furthermore, comparison with the non-implant contralateral limb suggested that rotation is not necessarily dictated by implant design.

1. Introduction

Total Knee Arthroplasty (TKA) utilises several different prosthesis designs to restore function in patients. Currently, designs such as the Posterior-Stabilising (PS) (fixed bearing component) and Cruciate-Retaining (CR) are characterised by a fixed motion, not allowing for a large rotational movement [1]. Research into the mechanical behaviour of the native knee in-vivo during functional movement has prompted some modification to the tibiofemoral articulation of TKA designs. One such design, the Medial

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Pivot (MP) implant, attempts to replicate the medial pivot motion of the native knee by modifying TKA joint geometry. While the concept of mimicking the human body in prosthetic designs is commendable, without a thorough approach and careful evidence-based progression in design there is a risk of compromising the function and longevity of the prosthesis.

The clinical results for MP prostheses in the peer-reviewed literature are somewhat mixed. Earlier studies have reported: concerns about limited post-operative Range of Motion (RoM) [2]; reasonably high incidence of complications (17%) and non-optimal component alignment [3]; and significantly lower post-operative clinical scores and increased self-reported pain compared to PS knees [4]. In contrast, Fan et al. [5] noted no failures or mal-alignment, and adequate RoM at an average of five years post-operatively. Vecchini et al. [6] described a low revision rate and comparable RoM. Lastly, a randomised control trial of staged bilateral TKRs found no significant differences in clinical scores or RoM between the MP and PS, PCL-retaining, bicruciate-retaining or mobile-bearing implants [7]. Although there may be no significant impact on clinical outcomes, does the MP design encourage more natural biomechanics, as the name may suggest?

It is still unknown whether knees replaced with the MP more reliably replicate tibiofemoral kinematics during walking than other existing implants. Studies have demonstrated comparable internal tibial rotation with weight-bearing knee flexion during walking [8]. Barnes et al. [9] reported a similar rotation pattern between implanted (MP) and intact knees. However, the magnitude of tibial rotation during mechanically-driven knee extension was significantly less in the implanted knees. Other in-vivo studies have demonstrated motion in the MP design that more closely resembles the physiological pattern in comparison to previous reports for PS and CR designs [10,11]. In contrast, Miyazaki et al. [12] reported an initial external tibial rotation followed by internal rotation in MP knees during a step-up task.

Regardless of any differences in motion, a more stable implant design should encourage muscle activity that more closely resembles a healthy or “physiological” knee. However, the evidence for the MP design in terms of improved extensor function over previous designs remains limited. Lester et al. [13] indicated that the knees replaced with a cruciate-retaining (CR) implant may use up to three times the muscle activity required to perform level walking compared to the non-operated contralateral limb up to two years post-arthroplasty. In contrast, Reynolds et al. [14] found no difference in quadriceps activation magnitude between MP knees and healthy age-matched controls during level or incline walking six months after total knee arthroplasty.

We hypothesised that the MP prosthesis design would restore more physiological knee function in-vivo, defined by an increase in the range of external rotation during the traditional screw-home mechanism, compared with PS and CR implant designs. We therefore aimed to investigate whether the MP knee in-vivo produces a more physiological behavior, as above, than the PS and CR implant types under the same task demands, by comparing tibiofemoral kinematics and muscular activity of the flexor and extensor muscles of the knee joint.

2. Material and methods

2.1. Cohort description

Patients with a Medial Pivot (Advance/Evolution; Microport Orthopaedics, TN, USA) (MP) implant were referred for inclusion by the treating consultant surgeon and recruited by a research assistant with face-to-face, email or phone contact. Patients were excluded if they had any other co-morbidities that may impair gait, such as an inadequate oxford score (Figure 1). Matching criteria (age, height, weight, operated limb) for each of the medial pivot patients were developed. This was provided to consulting surgeons who then returned a list of eligible PS (Legion; Smith&Nephew, UK) and CR (Nexgen; Zimmer Biomet, IN, USA) fixed-bearing implanted patients for recruitment by a research assistant. Patients were randomised and their group identification blinded from the research assistants performing the data collection. Ethical approval was granted from the local ethics committee prior to recruitment and each participant provided written informed consent prior to testing. A power analysis based on a repeated measures Analysis of Variance (ANOVA) (within factors: 1 group, 3 measurements, corr among repeated measure = 0.5) indicated that for a confidence of 0.95 (a = 0.05), an expected power of 0.95 and large effect size (f = 0.4), a minimum sample of 18 patients was required (G*power, v3.1.9.2).

From the final cohort of 18 matched patients in each group, eight patients had bilateral knee replacements resulting in a total sample of 26 implant knees for analysis in each group. Table 1 details the demographics, patient reported outcome measures (VR-12 and Oxford) and walking speeds of each group.

2.2. Motion analysis

Retroreflective markers were attached to anatomical landmarks of the feet, lower limbs and pelvis as per the Cleveland Clinic marker set (Figure 2a) [15], with additional markers on the 1st metatarsal, medial malleolus of the ankle, medial epicondyle of the knee, and opposing lateral iliac spines [16]. Marker positions were recorded using motion analysis consisting of 10 Bonita cameras and Nexus software® (Vicon, v2.6). A static calibration of the participant was performed to establish three-dimensional anatomical coordinate systems on the body segments, both in standing and supine positions.

Electromyography (EMG) data was collected using a Delsys Trigno® wireless EMG sensor system (Delsys Inc., USA). Patients were prepared for EMG electrode placement using the guidelines provided by the manufacturer. Sensors were placed over the muscle belly of five muscles (Figure 2b) to capture the activity of the quadriceps and hamstring muscle groups. The Delsys Trigno® system was connected to the Vicon Nexus® software to collect the EMG data simultaneously with the motion capture

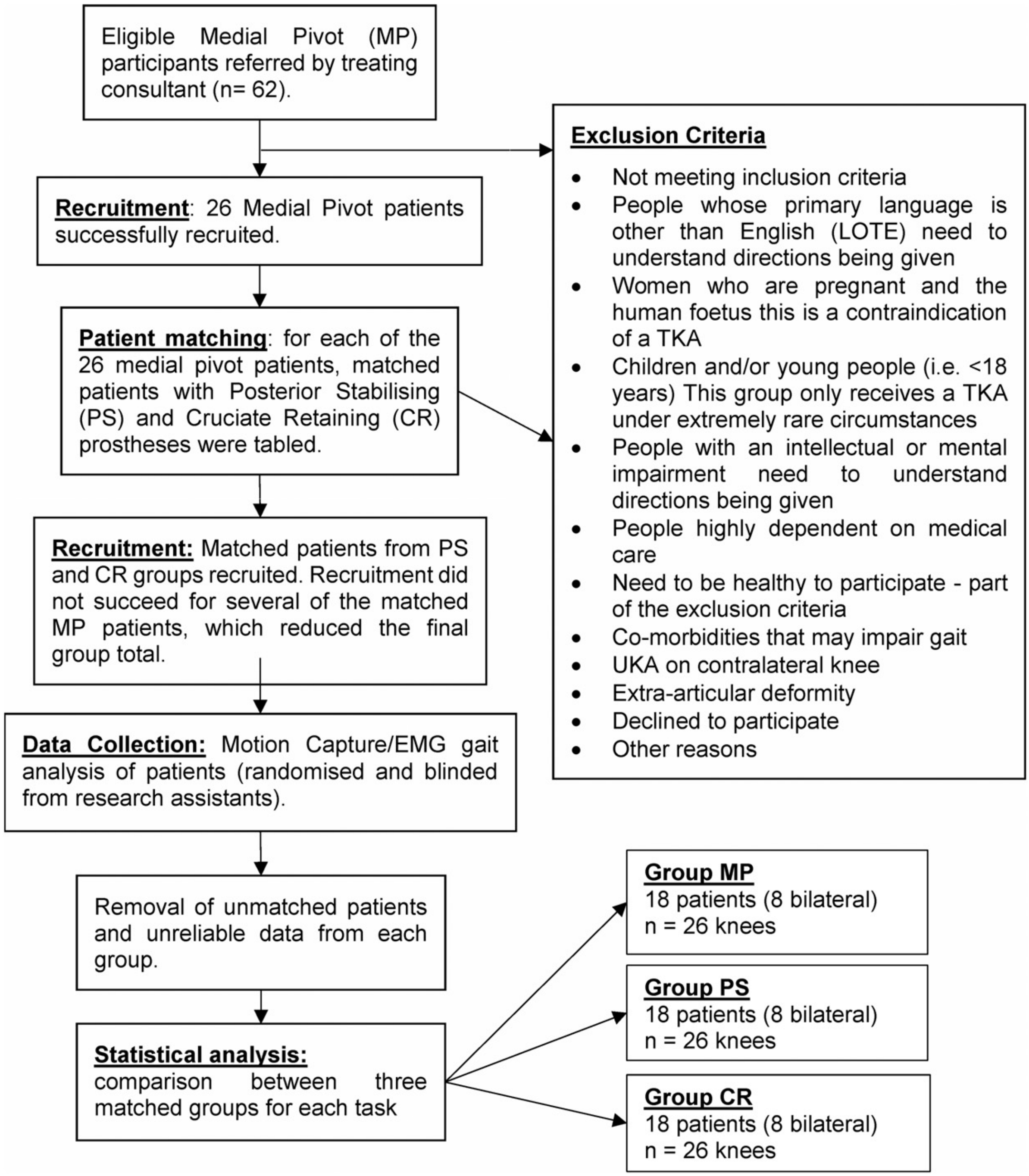

Figure 1. flowchart of study protocol, including: patient recruitment into Medial Pivot (MP), Posterior stabilising (PS), and Cruciate retaining (CR) groups; data collection; and statistical analysis.

data. The EMG data was collected at 1111 Hz and interpolated to 2000 Hz within the Nexus software, to match the 200 Hz motion capture data.

2.3. Task descriptions

Calibration of EMG data was conducted with Maximum Voluntary Isometric Contraction (MVIC) tests [17]. These were performed on seated weightlifting bench (with back support) by attaching an elastic rubber exercise strap (up to 60 kg load) from the bench to the patient's distal shank as resistance. Rest trials were taken with the patient relaxed and feet resting on the

Table 1
Mean ± SD [max–min] for group demographics, patient reported outcome measures and walking speeds (BMI: Body mass index).

Group	Medial pivot	Posterior stabilising	Cruciate retaining	P-value
Age (years)	66.7 ± 5.6 [57.8–80.4]	69.1 ± 6.4 [56.5–82.8]	67.6 ± 4.6 [58.4–77.1]	0.241
Height (cm)	166.6 ± 9 [154–179.5]	171.1. ± 9.3 [159.5–185.5]	169.6 ± 7.6 [156.5–179.5]	0.389
Weight (kg)	89.7 ± 15 [72–118]	89.8 ± 21.4 [60.8–148.8]	85.3 ± 15.7 [62.2–126]	0.736
BMI	32.3 ± 3.9 [24.4–39.3]	30.1 ± 5.4 [23.5–43.7]	29.5 ± 4.4 [23.7–40.2]	0.184
VR-12 Physical	46.2 ± 11.1 [22.6–55.7]	51.1 ± 5 [32.3–56]	51.2 ± 5.9 [34.6–56.6]	0.055
VR-12 Mental	58 ± 5.8 [46–65.7]	58.4 ± 3.4 [50.2–64.9]	57.9 ± 7.4 [32.4–64.4]	0.951
Oxford	45 ± 3.5 [35–48]	45.0 ± 3.2 [36–48]	45.1 ± 2.4 [40–48]	0.99
Comfortable walking speed (km/h)	2.6 ± 3.3 [1.5–4]	2.9 ± 0.7 [2–3.9]	3.0 ± 0.5 [2.2–4]	0.181
Fast walking speed (km/h)	3.3 ± 0.9 [2–5.2]	3.8 ± 0.9 [2.6–5.1]	3.8 ± 0.7 [2.9–5.2]	0.186

ground. Two exercises were conducted with the knees flexed at 90°: isometric knee extension to measure the quadriceps group, and isometric knee flexion to measure the hamstring group. These were performed as sustained contractions for two to three seconds, repeated three times with one-minute rest in between. Patients were verbally encouraged to push (or pull for the hamstring test) as hard as possible in an attempt to straighten (or bend for the hamstring test) the knee. Patients had several practices before completing a minimum of three recorded trials.

Patients then completed 10 step-ascents onto a stepping block (48 cm wide × 28 cm deep × 18 cm high) leading with the operated limb [16]. For the walking task, participants were familiarised for three to four minutes at each of the self-selected comfortable ('CWalk') and fast speeds ('FWalk') (130% of their comfortable speed) on an exercise treadmill (49 cm × 150 cm, TM435®, Vision Fitness Australia, Morwell Australia). Participants then walked for a total of three minutes at each of the respective speeds, with the middle 60 s used for analysis [16].

2.4. Data processing

Marker positions were imported from Nexus® to Visual3d® (v5, C-motion, USA) for calculation of joint kinematics. Marker positions were smoothed using a six-hertz Butterworth filter and assigned to a standard lower body model, established in the software. Joint kinematics were calculated and exported. The data was then input into MATLAB® (vR2016b, Mathworks, USA), where it was further processed to define the minimum, maximum and RoM during the phases of interest.

Within each MVIC trial, the highest EMG amplitude from 0.1 s moving average windows, calculated using the Root Mean Square (RMS) method, for each muscle was used to amplitude-normalise EMG data from the gait, in which an RMS during the phase of interest was calculated [18]. A co-contraction index (CCI) was also calculated to determine the simultaneous activation of antagonist muscles in the quadriceps and hamstring muscle groups [19], providing a measure of the relative stability across the lateral to medial sides of the knee joint. These included the relationship between Rectus Femoris and Biceps Femoris (RF-BF), Vastus Lateralis and Biceps Femoris (VL-BF), and Vastus Medialis and the Medial Hamstrings (VM-MH).

In the step-ascent task, the phase of interest was measured from when the lead leg made contact with the step and the supporting foot had left the ground, until the supporting leg contacted the step. For the walking tasks, the phase of interest was the terminal swing phase, defined as 87% to 100% of the gait cycle where terminal extension of the knee occurs [20].

2.5. Statistical analysis

A median of patient data (26 knees in each group) from all walking trials was calculated for use in statistical analyses, due to a lack of normal distribution. All statistical tests were performed in SPSS® (v20.0, IBM) with alpha set at <0.05. The normality and homoscedasticity of data were tested using the Shapiro–Wilk's and Levene tests, respectively. Analysis of kinematic and EMG results were performed using one-way repeated measures ANOVA (or Friedman's test for non-normally distributed data) within each task. Matched patients in the three implant groups were treated as the within subject repeated measures factor. Comparison within patients between the affected and unaffected contralateral knees (10 unilateral replacement patients per group) was performed with two-way repeated measures ANOVAs using two factors: knee (two levels) and time (three levels). Bonferroni adjustments were used for multiple comparisons. All results are presented as median (IQR) with a 95% confidence interval.

3. Results

3.1. Kinematics - extension

There were no significant differences between groups for the range of knee extension achieved in each of the step-ascent ($P = 0.33$), comfortable walk (CWalk) ($P = 0.24$), and faster walk (FWalk) ($P = 0.46$) tasks (Figure 3), and, on average, patients completed similar task demands, allowing equivalent assessment of kinematics and EMG for each group. Final extension values for each task are presented in Table 2. Patients in each group (MP, PS, and CR) finished all tasks in a significantly more flexed position than when standing ($P < 0.05$), and considerably more flexed than anatomical extension (0°).

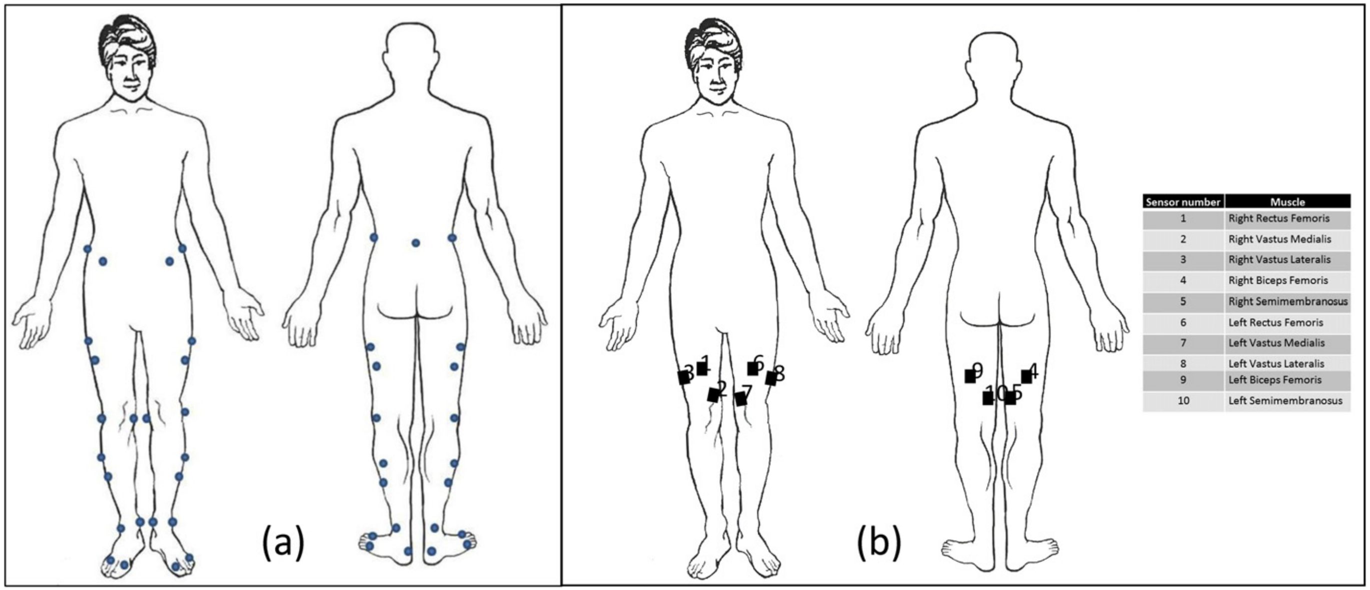

Figure 2. (a) Placement of markers, with the addition of the 1st metatarsal, medial malleolus of the ankle, medial epicondyle of the knee, and left and right iliac crests (used to reference the sacrum marker when the subject was in a supine position). (b) Placement of wireless EMG electrode sensors.

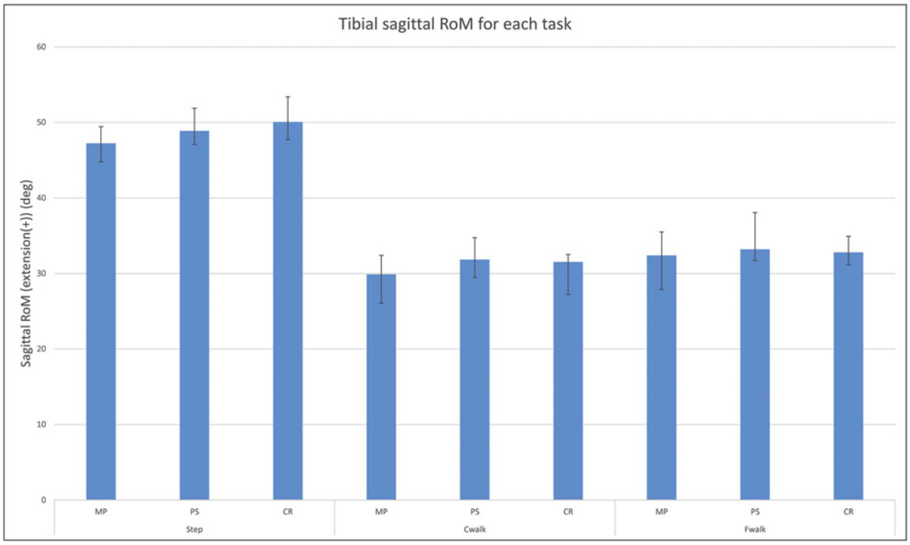

Figure 3. RoM of tibial flexion/extension for the three groups, across all the tasks during the respective phase of interest (Step = step-ascent, CWalk = comfortable self-selected walking speed, FWalk = 130% of CWalk, MP = Medial Pivot, PS = Posterior Stabilising, CR = Cruciate Retaining). A positive RoM value indicates an extension movement.

3.2. Kinematics — rotation

In all tasks, the three groups produced a paradoxical median internal rotation (Figure 4). There was a significant difference in the step-ascent task, between the MP (11.7°, 9.1–13.8) and CR implant groups (7.4°, 5–9.2) ($P = 0.007$), and between the PS (10.5°, 8.3–11.2) and CR groups (7.4°, 5–9.2) ($P = 0.01$). However, when the absolute values for rotation were analysed, to determine if there were any differences in the magnitude of rotation produced by each implant, there was only a significant difference between the MP and CR groups for the step-ascent task ($P = 0.011$). There were no significant differences between implant groups for the walking tasks.

Within each group, in each task, there was a large variance in the tibial rotation movement between patients. In the walking tasks, for example, all patients produced a similar range of knee extension motion during the terminal swing phase; however, the rotation movement did not display the same consistency. Several patients did not exclusively produce internal rotation, and some predominantly produced the desired external rotation. Furthermore, comparisons between the implant knee and non-implant contralateral knee in the unilateral patients ($n = 10$) resulted in no significant differences ($P > 0.05$), across each task.

3.3. Muscle activity

A comparison of EMG root mean square results, presented as % of MVIC for each muscle and group, is shown in Figure 5. There were no significant differences between the groups for any muscles, in any of the tasks, with one exception. The Vastus Lateralis muscle during the step-ascent task was significantly lower in activity in the MP group, compared with the CR group.

The CCI provides a measure of stability around the joint. There were, however, no significant differences between implant types for any co-contraction pairing, in any task (Figure 6). There was also no interaction effect between co-contraction pairings and groups. Although not significant, in all tasks, the VL-BF CCI was noticeably larger than the other muscle pairings.

4. Discussion

One of the main goals of TKR implant design has been to produce kinematics the same as, or at least similar to, the native knee, with the assumption that this will provide improved outcomes and patient satisfaction. Since the improved understanding

Table 2
Knee extension angle (mean [CI]) at the end of the phase of interest for each task. There was no significant difference between groups in any task; the p-value is reported in the final column. There was a significant difference between the standing and each of the other tasks, in all groups; the p-value is reported in the final row.

	MP	PS	CR	P-value
Standing	1.8° [−0.4, 5.2]	4.5° [2.5, 5.9]	3.9° [1.5, 7.9]	0.27
Step-ascent	12.6° [9.9, 14.7]	14.1° [11.8, 17.9]	15.2° [12.4, 18.6]	0.13
Comfortable Walk	14.1° [8.9, 15.8]	10.2° [5.8, 22.8]	9° [6.2, 12.7]	0.73
Faster Walk	11.1° [5.1, 16.1]	10.2° [5, 15.5]	7.8° [6.1, 11.7]	0.79
P-value (between Standing and each task)	<0.01	<0.01	<0.01	

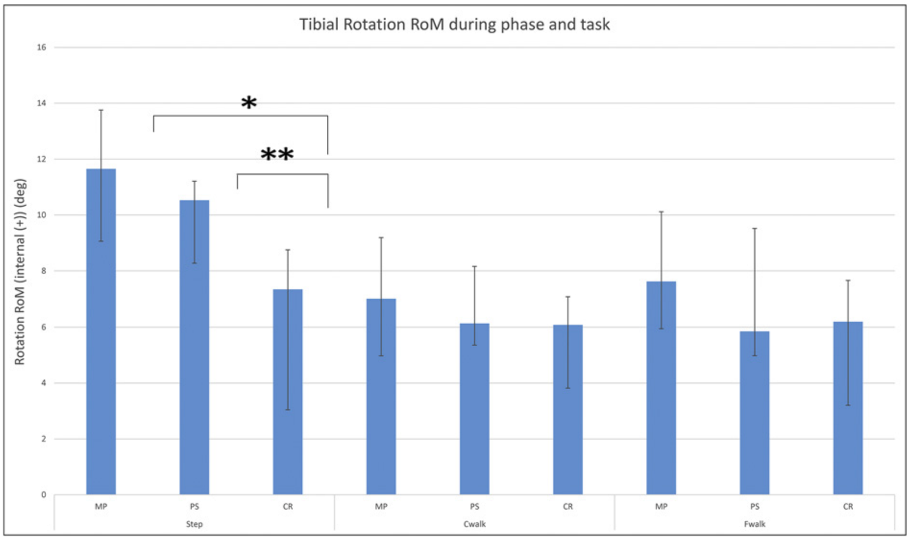

Figure 4. Tibial rotation ROM results for the three groups, across all tasks, during the phase of interest (terminal extension). * and ** denote significant differences between groups. The analysis of absolute rotation values resulted in a significant difference between the MP and CR groups in the step task only (*).

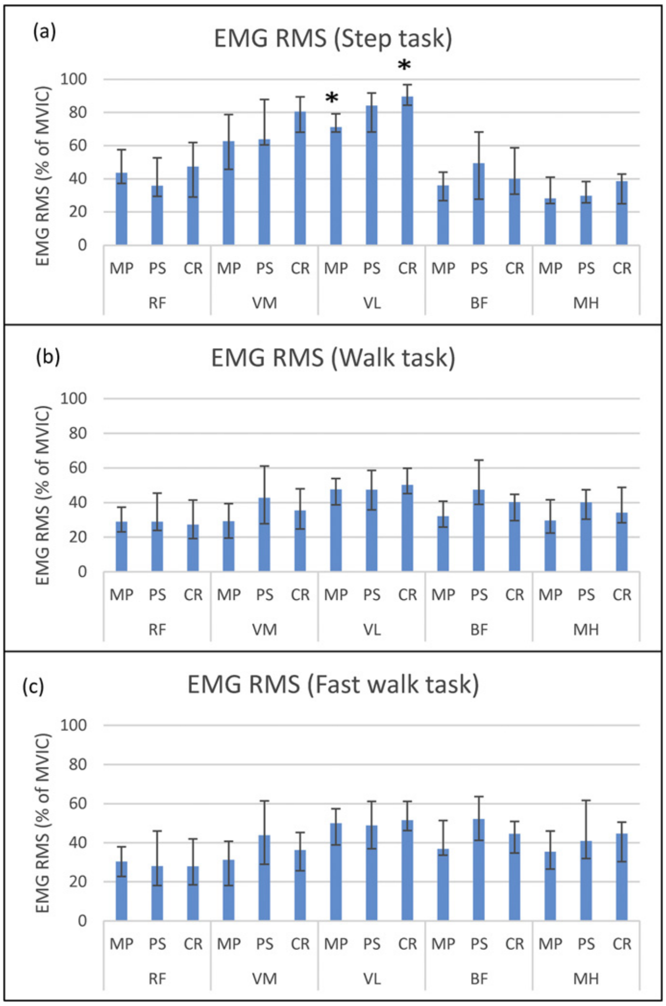

Figure 5. EMG RMS for step task (a), comfortable walking (b) and fast walking (c). (RF = Rectus Femoris, VM = Vastus Medialis, VL = Vastus Lateralis, BF = Biceps Femoris, MH = Medial Hamstrings).* denotes a significant difference between groups.

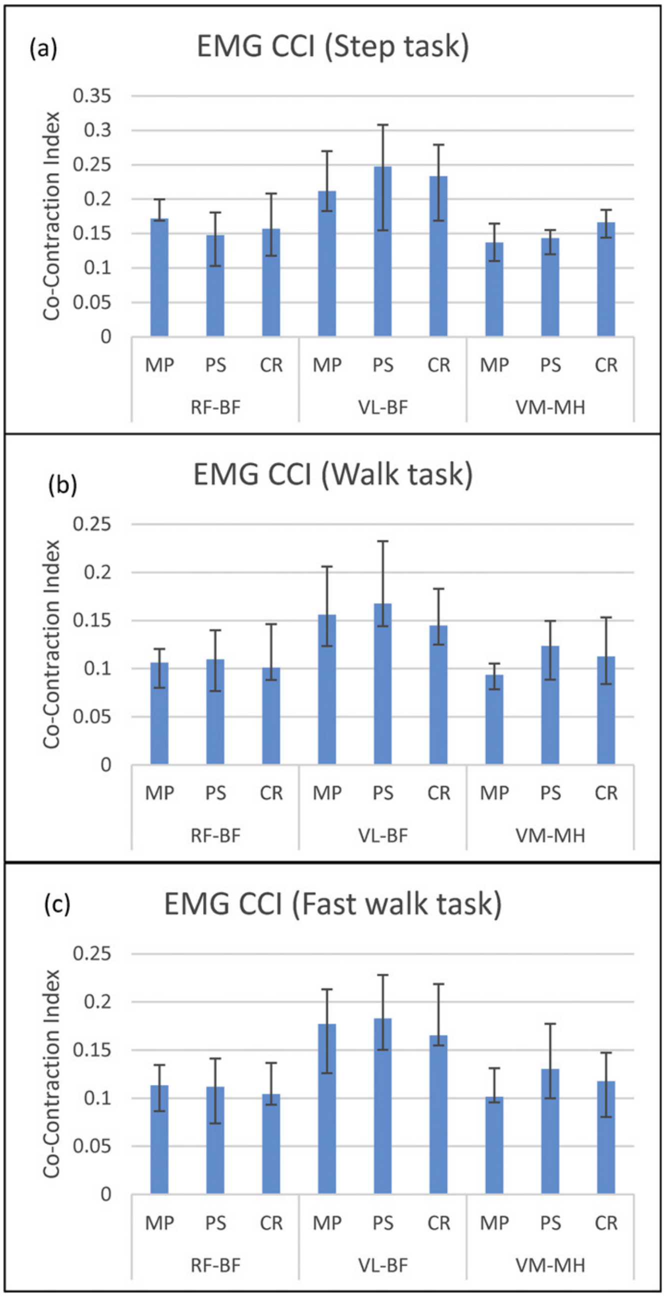

Figure 6. (a) Co contraction Indices for the step task, (b) comfortable walk and (c) fast walk. (RF-BF = Rectus Femoris to Biceps Femoris, VL-BF = Vastus Lateralis to Biceps Femoris, and VM-MH = Vastus Medialis to Medial Hamstrings).

of the usual kinematics of the native knee in the axial plane, with rotation described as occurring around a pivot point on the medial side, implant designs have been offered in an attempt to reproduce this. The "medial pivot" design is such a design with a highly congruent medial side and a less constrained lateral side, with the goal of reproducing this medial pivot and improving outcomes as a result. It is still somewhat controversial as to whether or not this design can actually reliably reproduce these desired kinematics, hence the purpose of this study was to specifically examine this aspect of kinematics in patients with well performing knee replacements doing specific functional tasks.

The results of this study did not confirm our hypothesis that the axial plane kinematics of the MP knee would more closely resemble a "physiological" knee, compared with fixed bearing knee implants (PS and CR). Whilst the MP knee did exhibit a greater range of absolute tibial motion compared with the CR knee during the step-ascent task, no differences were found for the walking tasks. However, the results also indicated that physiological mechanics were not achieved by any implant; a majority of patients in all groups produced a paradoxical internal rotation during terminal extension, in contrast to the physiological "screw-home" kinematics.

During the step-ascent task in this study, the medial pivot knee produced a significantly larger absolute range of rotational movement than the CR group. However, this was not apparent in the walking tasks (CWalk and FWalk). As such, the mechanical advantage of the MP design over other implants may be limited to certain conditions. More specifically, the walking tasks involved a lower range of tibial extension (Figure 3) and are categorised as open chain motion, whereas the step-ascent is a closed chain movement and consequently, may have invoked the different kinematic responses as the body pivots over the planted foot.

However, regardless of any differences between the designs, there was a large variation in rotation within each group during all tasks; assessed as a group, all designs produced an average internal rotation movement. This contradicts the traditional "screw-home" mechanism, defined as an anatomically and physiologically constrained movement of the knee involving an external rotation of the tibia during terminal extension [21]. Our finding is supported, partially, by a study which found PS and CR implant knees significantly lose the "screw-home" mechanism, compared with pre-op measurement, using a passive test with an intra-operative navigation system [22]. A lack of SHM has also been confirmed in the literature with an MP design [12]. This suggests that the pursuit of a more physiological design is more complex than a measurement of the range of rotational movement, given that all implant designs involve inverse kinematics.

This complication becomes even more challenging when presented with no significant difference between each implant and their contralateral native knee in this study, with the paradoxical internal rotation also evident in these physiological knees. Consequently, we cannot explicitly relate the paradoxical internal rotation nor the variation in rotation within each implant group to poor implant design. In support of this finding, Hallén and Lindahl [23] found that the direction of rotational movement of the tibia can be influenced by force, movement type, and the initial position of the tibia in flexion. The study also found that a conscious effort could be used to actively rotate the lower leg during the extension movement, either externally or internally. Comparably in our study, during the step-ascent task, patients may start with an open stance allowing for an external rotation of the tibia of the lead leg. They could then use momentum of the torso and contralateral limb to "swing" themselves onto the step, which would provide a passive internal rotation movement as mentioned by Hallén and Lindahl [23]. An explanation for these findings could be that patients are compensating with their healthy limb to create a symmetrical gait. Therefore, although we have found that it is possible to violate the "screw-home" mechanism without an implant, further investigation with a group of healthy controls during the same task demands may confirm the kinematics of a truly unaffected knee, without any confounding influences.

It is also possible that the results of either limb are influenced by patients not reaching full extension. Indeed, in all tasks, the knee was significantly more flexed than the respective standing extension angle ($P < 0.01$), and also much more flexed than full extension (0 degrees). If patients utilised the full extension available during the task, we may see different results. However, all patients did complete the task demands, with equivalent extension range between all groups, and therefore the results might be better viewed in this context. Particularly as the tasks were chosen to replicate two common activities of daily living, the PROMs results suggest that perhaps higher rotation is not required and as such, the potential differences in design are not of influence in these tasks.

In the absence of an overwhelming difference between implants and the discovery of paradoxical kinematics, differences in muscular activity may provide a better understanding for the benefits of the implant design. One suggested benefit of MP knees is improved stability, which we theorised may manifest as reduced muscle recruitment in this patient group. However, the only significant difference occurred between the MP and CR groups for Vastus Lateralis muscular activity in the step-ascent task (Figure 5). This could suggest that there is a reduced muscular demand associated with the MP knees, and thus corroborate previous findings in the literature which has suggested that CR implants may require up to three times the muscle activity during level walking compared to non-operated contralateral limb [13]. It is difficult for the authors to form a general conclusion; however, given the lack of other significant results in these findings when comparing between implants.

Furthermore, there were no significant differences in EMG results between the implant and contralateral native knees. This supports the findings of Reynolds et al. [14] who reported no difference in quadriceps activity between Advance/Evolution knees and healthy age-matched controls during level or incline walking six months after total knee arthroplasty. In addition, in all groups and tasks, there was generally a higher CCI in the VL-BF pairing (Figure 6), suggesting that more simultaneous muscular activity is required on the lateral side of the knee joint for stability, for all implant types. But again, a comparison with healthy controls would be necessary before concluding that this is due specifically to an implant.

The typical limitations involved with marker-based motion capture attached to the skin were apparent in this study; unwanted movement of the superficial tissue not related to movement of the bones is measured. These artefacts may have an influence on the results if not completely removed during the filtering of data. A bandage was wrapped around the thigh to secure the EMG sensors in place, but variances in the application of this bandage may have influenced the data collected. In addition, patients with a higher BMI and more subcutaneous tissue may produce a less reliable EMG signal. Performing a reliable MVIC test on older patients was also considered; perhaps performing simultaneous muscle dynamometry would provide verification of a patient's effort.

Due to the presence of bilateral implant patients in each group, the comparison between healthy and contralateral limbs had a reduced sample size of 10 patients per group. In addition, although patients' contralateral knees were asymptomatic and deemed not to require at TKA, there is no guarantee that these resemble a truly healthy knee void of any condition that may influence gait. Therefore, a healthy control group may provide a more detailed understanding of physiological knee motion during the tasks, and strengthen the conclusions of the study regarding the validity of the natural "screw-home" mechanism. Furthermore, we analysed the terminal extension phase in each task; however, other phases of movement may provide more significant and clinically relevant differences between the implant types. For instance, during the loading response phase of gait when the knee joint starts to flex, Kim et al. [21] have reported a "paradoxical" screw-home mechanism, where the tibia externally rotates, rather than internally rotates as anticipated. It would be interesting to investigate whether this applies equally in some or all implant types.

5. Conclusions

Although the MP implant is designed to be produced more physiological kinematics, the design only exhibited a greater range of tibial rotational motion than the CR knee during one specific task of step-ascent. Furthermore, none of the implant designs exhibited a physiological "screw-home" mechanism defined by an external rotation during terminal extension, and all designs generally involved similar muscular activity. There were significant variations between patients in both TKR and native knees, and kinematics did not appear to be specifically driven by implant design. It is important to note that all patients in this study were happy patients with good outcomes, and therefore it would seem that production of presumed physiological kinematics is not required for a good outcome, and kinematics promised by specific implant designs are certainly not consistently delivered.

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References

[1] Kolisek FR, McGrath MS, Marker DR, Jessup N, Seyler TM, Mont MA, et al. Posterior-stabilised versus posterior cruciate ligament-retaining total knee arthroplasty. Iowa Orthop J 2009;29:23–7.
[2] Shakespeare D, Ledger M, Kinzel V. Flexion after total knee replacement. A comparison between the Medial Pivot knee and a posterior stabilised implant. Knee 2006;13:371–3.
[3] Karachalios T, Roidis N, Giotikas D, Bargiotas K, Varitimidis S, Malizos KN. A mid-term clinical outcome study of the Advance Medial Pivot knee arthroplasty. Knee 2009;16:484–8.
[4] Kim YH, Yoon SH, Kim JS. Early outcome of TKA with a medial pivot fixed-bearing prosthesis is worse than with a PFC mobile-bearing prosthesis. Clin Orthop Relat Res 2009;467:493–503.
[5] Fan CY, Hsieh JT, Hsieh MS, Shih YC, Lee CH. Primitive results after medial-pivot knee arthroplasties: a minimum 5-year follow-up study. J Arthroplasty 2010;25:492–6.
[6] Vecchini E, Christodoulidis A, Magnan B, Ricci M, Regis D, Bartolozzi P. Clinical and radiologic outcomes of total knee arthroplasty using the Advance Medial Pivot prosthesis. A mean 7 years follow-up. Knee 2012;19:851–5.
[7] Pritchett JW. Patients prefer a bicruciate-retaining or the medial pivot total knee prosthesis. J Arthroplasty 2011;26:224–8.
[8] Schmidt R, Komistek RD, Blaha JD, Penenberg BL, Maloney WJ. Fluoroscopic analyses of cruciate-retaining and medial pivot knee implants. Clin Orthop Relat Res 2003:139–47.
[9] Barnes CL, Blaha JD, DeBoer D, Stemniski P, Obert R, Carroll M. Assessment of a medial pivot total knee arthroplasty design in a cadaveric knee extension test model. J Arthroplasty 2012;27:1460–8.
[10] Ishida K, Matsumoto T, Tsumura N, Chinzei N, Kitagawa A, Kubo S, et al. In vivo comparisons of patellofemoral kinematics before and after ADVANCE Medial-Pivot total knee arthroplasty. Int Orthop 2012;36:2073–7.
[11] Kitagawa A, Ishida K, Chin T, Tsumura N, Iguchi T. Partial restoration of knee kinematics in severe valgus deformity using the medial-pivot total knee arthroplasty. Knee Surg Sports Traumatol Arthrosc 2012;22(7):1599–606.
[12] Miyazaki Y, Nakamura T, Kogame K, Saito M, Yamamoto K, Suguro T. Analysis of the kinematics of total knee prostheses with a medial pivot design. J Arthroplasty 2011;26:1038–44.
[13] Lester DK, Shantharam R, Zhang K. Dynamic electromyography after cruciate-retaining total knee arthroplasty revealed a threefold quadriceps demand compared with the contralateral normal knee. J Arthroplasty 2013;28:557–62.
[14] Reynolds S, Dervin G, Mario L. Lower limb kinetics and quadriceps activation following total knee arthroplasty with the medial pivot knee during level and incline walking. 30th Annual Conference of Biomechanics in Sports. Melbourne, Australia; 2012.
[15] Scholes CJ, Lynch JT, Ebrahimi M, Fritsch BA, Parker DA. Gait adaptations following multiple-ligament knee reconstruction occur with altered knee kinematics during level walking. Knee Surg Sports Traumatol Arthrosc 2016;25(5):1489–99.
[16] Beach A, Scholes C, Parker D. The effect of total knee arthroplasty on active knee extension during treadmill walking. Asia-Pac J Sports Med Arthrosc Rehabil Technol 2016;6:38.
[17] Ushiyama N, Kurobe Y, Momose K. Validity of maximal isometric knee extension strength measurements obtained via belt-stabilised hand-held dynamometry in healthy adults. J Phys Ther Sci 2017;29:1987–92.
[18] Hubley-Kozey CL, Deluzio KJ, Landry SC, McNutt JS, Stanish WD. Neuromuscular alterations during walking in persons with moderate knee osteoarthritis. J Electromyogr Kinesiol 2006;16:365–78.
[19] Lewek MD, Rudolph KS, Snyder-Mackler L. Control of frontal plane knee laxity during gait in patients with medial compartment knee osteoarthritis. Osteoarthr Cartil 2004;12:745–51.
[20] Simoneau G. Kinesiology of walking. In: Neumann DA, editor. Kinesiology of the Musculoskeletal System: Foundations for Rehabilitation, 2. St Louis: Mosby; 2010. p. 636.
[21] Kim HY, Kim KJ, Yang DS, Jeung SW, Choi HG, Choy WS. Screw-home movement of the tibiofemoral joint during normal gait: three-dimensional analysis. Clin Orthop Surg 2015;7:303–9.
[22] Collins DJ, Khatib YH, Parker DA, Jenkin DE, Molnar RB. Tibial rotation kinematics subsequent to knee arthroplasty. J Orthop 2015;12:7–10.
[23] Hallén LG, Lindahl O. The "screw-home" movement in the knee-joint. Acta Orthop Scand 1966;37:97–106.