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modification program (Table 1). This was associated with an average 6.7 degree increase in self-selected toe-out angle ($p < 0.001$) and 10% reduction in the late stance KAM ($p = 0.04$). Participants reported that difficulty in achieving the desired toe-out angle significantly decreased over the course of the program. Joint discomfort was reported by five participants (33%) in the hip or knee joints, though none lasted longer than two weeks.
Conclusion: Results from the current study provide preliminary evidence as to the benefits of toe-out gait modification training in individuals with medial knee OA. Specifically, these findings suggest that gait modification can significantly improve clinical and biomechanical outcomes relevant to medial compartment knee OA. This study also showed that gait modification can be successfully delivered with minimal difficulty or consequences to other lower limb joints. Future research utilizing more participants and a control group are now needed to best understand the biomechanical and clinical changes following toe-out gait modification.

Table 1
Mean (SD) outcome measures

	Baseline (Week 0)	Follow-up (Week 11)	p-value
Biomechanical outcomes			
Self-selected toe-out angle (o)	4.75 (6.59)	11.41 (6.46) *	<0.001
Early stance peak KAM (%BW*ht)	3.45 (0.82)	3.17 (0.72)	0.12
Late stance peak KAM (%BW*ht)	2.87 (0.92)	2.63 (0.84) *	0.04
KAM impulse (%BW*ht*sec)	1.33 (0.29)	1.24 (0.34)	0.20
Clinical outcomes			
WOMAC pain subscale (0-20)	7.4 (3.4)	5.3 (2.9) *	0.02
WOMAC total score (0-96)	36.9 (14.8)	26.4 (13.5) *	0.02
NRS pain (0-10)	4.5 (1.7)	2.6 (1.8) *	<0.001

*Indicates significant difference ($p < 0.05$).

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MUSCLE ACTIVATION PATTERNS FOLLOWING KNEE JOINT REPLACEMENT DURING THE POSTURAL CONTROL TASK

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Purpose: The total knee arthroplasty (TKA) is performed for the purpose of pain reduction and functional improvement for patients with knee osteoarthritis. TKA patients have demonstrated reduced walking velocity and stride length during walking. This performance deficit may emanate from impaired balance control resulting from changes in muscle activity. It was reported that patients with ankle instability showed muscle activation patterns different from healthy subjects and that muscle activation patterns of the asymptomatic leg changed than the symptomatic side in patients with patellofemoral pain. However, none of the studies investigated muscle activation patterns of TKA patients during the postural control task. The purpose of this study was to examine the muscle activation patterns of the TKA patients during task of transition from double leg standing position to single leg standing position.
Methods: Ten subjects (two men, eight women, mean age 68.9 $\pm$ 6.0 years) that four weeks passed after TKA and ten healthy, age-matched control participants (healty group: one man, nine women, mean age 68.0 $\pm$ 5.7 years) participated in this study. Each participant provided informed consent to the potential risks associated with their participation. Subjects performed a transition task from double leg standing position to single leg standing position after a beep sound from a electromyograph (EMG) immediately, and surface EMG signals were recorded using a 8-channel electrode system (Myosystem 1400) at the time. The baseline EMG was calculated by averaging the EMG activity for :100ms interval in a resting position. The onset of EMG activity of each muscle was determined when the EMG amplitude exceeded two standard deviations of the baseline level from a beep sound time. EMG signals of the following muscles were recorded at the both legs in TKA patients and at the dominant leg in control subjects: gluteus maximus, gluteus medius, adductor longus, vastus lateralis, biceps femoris, tibialis anterior and lateral gastrocnemius. Also, Foot switches were attached to the sole of the lower limbs which made elevation done and measured the motor reaction time when a foot left the floor. All date were analyzed for the supporting leg only and were averaged across three trials. Two-way analysis of variance was used to compare the onset of muscle activity between groups and compare the onset of muscle activity of each muscle with the motor reaction time in each group. We used the Bonferroni method as an adjustment of the multiple comparisons. The level of significance was set at 0.05.
Results: In the comparison between groups, the onset of muscle activity of the vastus lateralis was significantly later in the operation side (0.76 $\pm$ 0.36s) and the non-operation side (0.77 $\pm$ 0.47s) than the healthy group (0.44 $\pm$ 0.12s). There was no significant difference between groups in other muscle activity onset. In the comparison in each group, the entire onset of muscle activity except the vastus lateralis was significantly earlier than the motor reaction time in the operation side of the TKA subjects. The onset of muscle activity of gluteus medius, adductor longus and the tibialis anterior were significantly earlier than the motor reaction time in the non-operation side. In the healthy group, onset of muscle activity of gluteus medius muscle, adductor longus, vastus lateralis, biceps femoris and the tibialis anterior were significantly earlier than the motor reaction time.
Conclusions: The results of this study showed that the onset of muscle activity of the vastus lateralis in the operation side and the non-operation side of TKA subjects was later than the healthy group. Also, the muscle that the onset of muscle activity was earlier than the motor reaction time was different from the healthy group in the operation side and the non-operation side of TKA subjects. Horak et al described that the subjects used the posture control strategy due to the hip joint as compensation, when they reduced the somatosensory of the ankle of subjects. We suggest that the TKA patients may use the muscle activation patterns different from a healthy subject compensating the myofunction of the quadriceps femoris in both operation side and non-operation side during postural control task. Because the difference in observed muscle activity pattern may influence posture control after TKA, a further investigation is necessary.

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PARTIAL WEIGHT BEARING AND CONTINUOUS PASSIVE MOTION FOR REHABILITATION FOLLOWING MICROFRACTURE SURGERY: A MULTISCALE FINITE ELEMENT SIMULATION

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Purpose: Continuous passive motion (CPM) and partial weight bearing (PWB) are often applied in rehabilitation after microfracture surgery for cartilage repair. CPM is often used immediately after surgery and is thought to provide lower mechanical loads than PWB. In this study, using a multi-scale finite element analysis (FEA) and an in vitro microfracture system, we propose to compare 1) the mechanical loading profiles at lesion site during PWB and CPM, and 2) the biophysical fields, including stress, strain, fluid pressure, and nutrient transport, in both repaired tissue and surrounding cartilage during PWB and CPM.
Methods: PWB and CPM movements were simulated adopting the OpenKnee model in FEBio, based on the MRI imaging of a healthy female knee joint. PWB was modeled as vertical load of 0-400N ($\sim$1/2 donor body weight) on the femur (tibia was fixed) while the CPM as a 0-30° flexion, both in quasi-static conditions. The deformation and stress within 20 different cartilage regions (Fig 2C) on the medial condyle head were determined. The cartilage deformation field was further used as input into a tissue scale model (Fig 1C&D), in which the cartilage lesion and surrounding tissue were modeled as nonlinear biphasic materials with strain dependent permeability. The material properties of the lesion were obtained by testing the repaired tissue from an in vitro organ culture model (Fig 1E&F). Microfracture was simulated on a bone-cartilage explant, which was cultured with mechanical stimulation for two months. Multiple simulations were performed to compare 1) the effects of PWB and CPM, 2) effect of the lesion size (6-10 mm), and 3) effect of material properties at lesion site to simulate the "healing phase" (from very soft marrow clot to fibrous cartilage).

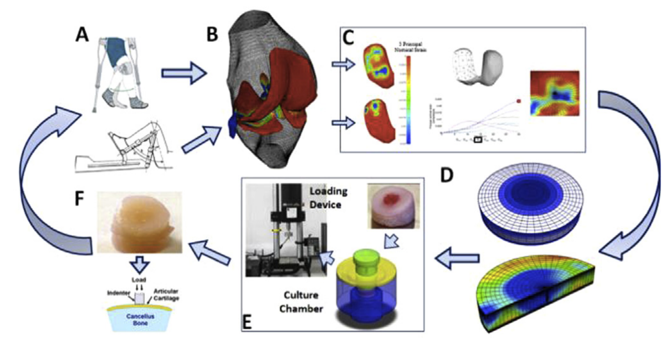

Figure 1. Illustration of overall research design. (A) Knee loading profiles during PWB and CPM are extracted from physical therapy studies and input into a 3D FEM knee joint model. (B-C) Set up a 3D FEM knee model to calculate the load distribution on medial condyle cartilage. (D) Biomechanical fields in the lesion area and surrounding healthy cartilage, including stress, strain, fluid pressure and flow, are predicted using a FEM nonlinear inhomogeneous biphasic model. (E) An in vitro microfracture model which can adopt the loading profiles determined by FEM simulation. (F) New cartilage tissue with functional stiffness fully filled the cartilage lesion site and evaluated with indentation.

Results: Cartilage deformation at different regions of the medial condyle were calculated (Fig 2A&B). When $^1/_4$ body weight is applied on the knee, cartilage deformation locates mainly in the proximal and distal regions (∼2-3% strain) (Fig 2A). During CPM, the highest deformation appears solely in the distal region (Fig 2B). Interestingly, the highest cartilage deformation during 30° flexion (Fig 2C region Q15 ∼2.5%) is comparable to that of $^1/_4$ PWB (Fig 2D&E).

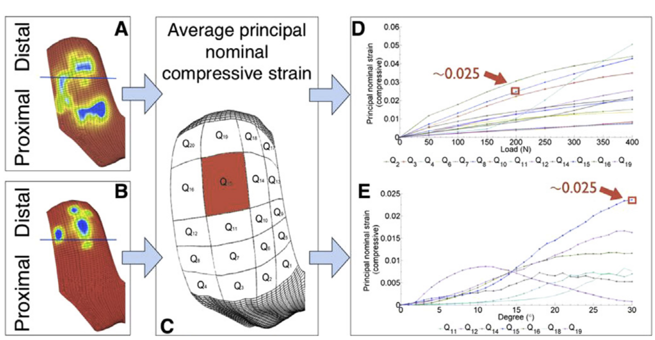

Figure 2. Principal nominal compressive strains in PWB (A&D) and CPM (B&E). The contour plots show a concentration in the distal region for both cases. The trend of the principal nominal compressive strains during PWB (D) highlights that, at 200 N loading, the strain magnitude is comparable with CPM (E) in region Q_{15} (red arrows).

In the proximal region, PWB generates much higher compression at the lesion site than CPM. The compressive strain at the lesion site during CPM is twice that of PWB (Fig 3A). On the contrary, the total stress at lesion site is higher for PWB (Fig 3B). When the lesion is at the distal region, CPM might generate even higher deformation within the lesion site and higher total stress within the surrounding tissue than PWB, i.e. CPM does not necessarily imply more moderate stimulations than PWB. PWB shows higher fluid flux (index of nutrient transport) concentration at deep zone than CPM (Fig 3E). The fluid load support is ∼93% for the lesion and ∼76% for the surrounding tissue, for both PWB and CPM. Increase of lesion diameter decreases the fluid load support by ∼8% in the surrounding cartilage, which implies higher stress and friction in cartilage solid matrix. During the "healing phase" the fluid load support decreases by ∼17% within the lesion. Contrarily, the surrounding cartilage shows a constant fluid support ratio of ∼76% with no relation to lesion stiffness.

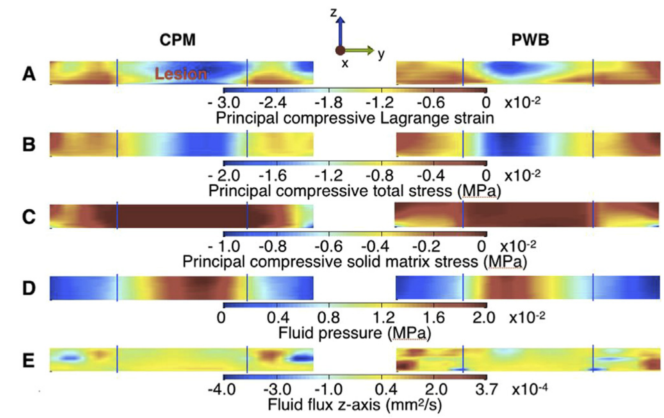

Figure 3. Physical fields show that PWB and CPM load the tissue in substantial different manners. The compressive strain (A) distribute more evenly under CPM. Stress at lesion site (B&C&D) is higher under PWB. Fluid flux at lesion boundary is also more efficient under PWB (E).

Conclusions: This work, for the first time, compared mechanical stimulations on cartilage at different regions of the medial condyle during CPM and PWB post microfracture surgery. At distal region on medial condyle, CPM provides a larger mechanical stimulation than PWB in terms of strain magnitude and distribution. Mechanical stimulation on repaired tissue and surrounding cartilage also changes with the lesion size and lesion stiffness.

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A QUANTITATIVE ASSESSMENT OF VARUS THRUST IN PATIENTS WITH MEDIAL OSTEOARTHRITIS

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Purpose: Varus thrust is an abnormal lateral knee motion frequently seen with medial OA patients, occurring in the early part of the stance phase. Lower limb alignment influences how joint loads are distributed between the medial and the lateral compartments and is considered to be crucial to the diseases' prognosis. The varus thrust is a worsening of the alignment in the stance phase of the gait cycle and is known to closely relate to disease progression. However, the thrust during walking has not been well examined and no previous studies have analyzed it quantitatively. The purpose of this study was to measure the varus thrust quantitatively and to examine the relationship with other dynamic (knee adduction moment), static evaluations (femoro-tibial angle in weight bearing X-P) and OA grade.
Methods: Forty-four knees in 32 patients (mean age, 72 years; range, 64-81 years) who exhibited the radiographic OA at least grade 2 according to the Kellgren-Lawrence (K-L) scale were enrolled. Six retroreflective markers were directly placed onto the skin of the affected limb. To simplify the measurement of the thrust, we used skin markers placed on the greater trochanter, lateral joint line of the knee, and lateral malleolus. The marker hip-knee-ankle angle (HKA angle) was represented by the angle formed by the three markers on the coronal plane (Fig. 1). Typically, medial OA knees have a peak in varus angle on the coronal plane in the initial stance phase corresponding to the thrust motion; therefore, the amount of thrust motion was defined by differences in the marker HKA angles between heel strike and the first varus peak (Fig.2). Knee kinematics and kinetics during the gait were measured using a Pro-reflex three-camera system (120 frames/s), and an AM6110 force plate (frequency 600 Hz, sample frequency synchronized to 120 Hz) to obtain the ground reaction force during motion. Three-dimensional knee kinetics were then assessed using an inverse dynamics approach and knee adduction moments were normalized to percent body weight times height (%BW*Ht).
Results: As a whole, the marker HKA angle at heel strike was 184.1°(±7.5°) and increased to 187.3°(±6.4°) at the first varus peak of the initial stance phase, and the amount of thrust was 3.2°(±2.7°). The