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How symmetric are metal-on-metal hip resurfacing patients during gait? Insights for the rehabilitation

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A B S T R A C T

Metal-on-metal hip resurfacing patients demonstrate hip biomechanics closer to normal in comparison to total hip arthroplasty during gait. However, it is not clear how symmetric is the gait of hip resurfacing patients. Biomechanical data of 12 unilateral metal-on-metal hip resurfacing participants were collected during gait at a mean time of 45 months (SD 24) after surgery. Ankle, knee, hip, pelvis and trunk kinematics and kinetics of both sides were measured with a motion and force-capture system. Principal component analysis and mean hypothesis' tests were used to compare the operated and healthy sides. The operated side had prolonged ankle eversion angle during late stance and delayed increased ankle inversion angle during early swing ($p = 0.008$; effect size = 0.70), increased ankle inversion moment during late stance ($p = 0.001$; effect size = 0.78), increased knee adduction angle during swing ($p = 0.044$; effect size = 0.57), decreased knee abduction moment during stance ($p = 0.05$; effect size = 0.40), decreased hip range of motion in the sagittal plane ($p = 0.046$; effect size = 0.56), decreased range of hip abduction moment during stance ($p = 0.02$; effect size = 0.63), increased hip range of motion in the transverse plane ($p = 0.02$; effect size = 0.62), decreased hip internal rotation moment during the transition from loading response to midstance ($p = 0.001$; effect size = 0.81) and increased trunk ipsilateral lean ($p = 0.03$; effect size = 0.60). Therefore, hip resurfacing patients have some degree of asymmetry in long term, which may be related to hip weakness and decreased range of motion, to foot misalignments and to strategies implemented to reduce loading on the operated hip. Interventions such as muscle strengthening and stretching, insoles and gait feedback training may help improving symmetry following hip resurfacing.

1. Introduction

Total hip arthroplasty (THA) is a treatment option for end-stage hip failure, with a cost of $38,295 in the United States (International Federation of Health Plans, 2013). Because of the low durability and longevity of the prosthesis (Sharkey et al., 2006), THA is not promising for more active patients (Crowninshield et al., 2006). In fact, more candidates for THA have become physically active (Crowninshield et al., 2006). Metal-on-metal hip resurfacing arthroplasty (hereafter referred to as hip resurfacing) is an alternative to THA for active patients (Hing et al., 2007), since it is bone-conservative, has shorter recovery time and reduces the implant dislocation risk (Mehra et al., 2015; Pollard, 2006). Longitudinal studies have demonstrated favorable results for hip resurfacing in comparison to THA regarding aseptic loosening, stability, toxicity of wear and implant survivorship (Azam et al., 2016; Australian Orthopedic Association, 2015). In addition, hip resurfacing allows patients to return unrestrictedly to their activities within a year of the procedure (Pollard, 2006), which has driven younger patients to request hip resurfacing as an alternative to THA (Pollard, 2006). In Canada, between 2009 and 2014, there was an increase of 11.8% in the number of hip resurfacing (Canadian Joint Replacement Registry, 2015).

It is speculated that, in comparison to THA, hip resurfacing contributes to greater weight bearing on the operated side during activities such as walking (Aqil et al., 2013), which may be

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explained by the larger femoral head sizes used with the technique. Increased body weight bearing may help to explain the hip extension range of motion and abduction moments closer to normal as demonstrated after hip resurfacing in comparison to THA (Mont et al., 2007). However, it is not clear if the biomechanics of the lower limb with hip resurfacing are similar to the biomechanics of the contralateral lower limb. It is possible that individuals with unilateral hip resurfacing still have asymmetric gait patterns after surgery (Mellon et al., 2014), which may overload the contralateral side. For example, after THA, 79% of the patients developed osteoarthritis in the contralateral hip and 54% had undergone another arthroplasty (Ritter et al., 1996). In addition, asymmetric mechanics in other joints might be expected. For example, it is possible that hip resurfacing individuals increase trunk ipsilateral lean during gait to laterally shift the body center of mass and consequently minimize the hip abduction moment on the operated side. Although this strategy may reduce the load on the operated hip, it may overload the spine joints, such as the intervertebral and facet joints (Popovich et al., 2013).

This study compared the biomechanics of the operated side of individuals with unilateral hip resurfacing with the biomechanics of the contralateral side during gait. It was hypothesized decreased hip abduction and extension moments along with decreased hip extension angle and increased ipsilateral trunk lean during the stance phase in the operated side.

2. Material and methods

2.1. Participants

Sample size was determined using the software G*Power (Faul et al., 2007) with the following input data: two-tailed dependent *t*-test, statistical power of 80%, significance level of 0.05, and the mean effect size of the differences in hip flexion-extension angle ($d = 0.65$), adduction-abduction moment ($d = 0.79$) and internal-external rotation moment ($d = 1.31$) found in a pilot study with 5 subjects ($d = 0.92$). This resulted in an estimated sample size of 12 participants. Twenty-three potential participants were invited to participate in the study, but 11 did not want to take part in the study or did not meet the inclusion criteria. Therefore, ten males and two females with unilateral metal-on-metal hip resurfacing participated in the study. The senior author was responsible for the hip resurfacing surgery in all participants using a direct lateral approach to the hip joint, with the specific inclusion and exclusion criteria for the surgery described in a previous study (Bow et al., 2012). All implants were the Depuy Orthopaedic ASR (Warsaw, Ind). The surgical technique has been previously described in detail (Bow et al., 2012; Kunz et al., 2010). Patients were allowed to fully weight bear soon after surgery. Mobilization with physiotherapy began within 24 h of surgery and continued until the patient was discharged, usually within 2–3 days of surgery. A non-supervised home exercise program was provided to the patients on discharge to continue to improve their strength and range of motion about the hip joint.

The inclusion criteria were a minimum of 12 months of follow-up after the surgery, no history of falls and no other surgeries or injury to either lower limbs in the past twelve months, no history of stroke or any other form of arthritis, neuromuscular or cardiovascular disorders, being able to walk without assistive device and a city block, and to climb stairs in a reciprocal fashion. The exclusion criterion was the report of pain or walking unsteadily during data collection. Each participant signed a consent form approved by the university's Ethical Research Committee.

2.2. Procedures

The participants answered the Western Ontario and MacMaster Universities Osteoarthritis Index (WOMAC) and the Lower Extremity Activity Scale (LEAS). The WOMAC is validated for evaluating outcome after THA (Bellamy et al., 1988), with scores varying from 0 to 100 and higher scores indicating better condition in the pain, stiffness and function dimensions. The LEAS is validated for the assessment of patients' actual activity levels (Saleh et al., 2005), with scores varying from 1 to 18 and higher scores indicating higher activity level. Then, the heights and masses of the participants were measured. Subsequently, gait data were recorded at 200 Hz using a 12-camera motion capture system (Oqus 4, Qualisys, Gothenburg, Sweden) synchronized with six force platforms (Custom BP model, AMTI, Massachusetts, USA). The force platforms registered ground reaction force data at a frequency of 1000 Hz, which was subsequently downsampled at 200 Hz.

Anatomical and clusters of tracking markers were used to determine the coordinates of the trunk, pelvis, thigh, shank and feet (Cappozzo et al., 1995) using data obtained with the participant in a relaxed standing position (static trials). Participants then walked at their self-selected speed wearing their own shoes for five trials along a 15-m distance (Fig. 1).

2.3. Data reduction

Gait data were processed using the Visual3D (C-motion, Inc., Rockville, USA). Raw kinematic and force data were filtered using a low-pass fourth order Butterworth filter with a cut-off frequency set at 6 Hz and 18 Hz, respectively. Heel contact and toe-off were determined automatically in Visual3D using the vertical ground reaction force at threshold of 20 N. The following joint kinematics were calculated: (1) ankle dorsiflexion-plantar flexion (medio-lateral axis), inversion-eversion (antero-posterior axis) and adduction-abduction (longitudinal axis) with respect to the shank; (2) knee flexion-extension, adduction-abduction and internal-external rotation, represented by the motion of the shank relative to the thigh; (3) hip flexion-extension, adduction-abduction and internal-external rotation, represented by the motion of the thigh relative to the pelvis; (4) pelvic anteversion-retroversion (medio-lateral axis), ipsilateral-contralateral drop (antero-posterior axis) and external-internal rotation (longitudinal axis) with respect to the lab; (5) trunk flexion-extension (medio-lateral axis), ipsilateral-contralateral lean (antero-posterior axis) and external-internal rotation (longitudinal axis) with respect to the lab. Kinetic data included ankle, knee and hip internal moments in the sagittal, frontal and transverse planes. Both kinematic and kinetic data were computed in the joint coordinate system (Grood and Suntay, 1983). Joint moments were calculated using the inverse dynamic approach, normalized to body mass (kg), and reported in Nm/kg. Internal joint moments were reported throughout the text. Kinematics and kinetics data were normalized to 101 data points, one for each percentage of the gait cycle.

2.4. Data analysis

2.4.1. Principal component analysis (PCA)

Extracting discrete variables from temporal series has at least four limitations: (i) severe data reduction, (ii) loss of temporal information, (iii) difficulty to define the parameter to extract and (iv) high correlation between the extracted discrete variables (i.e. redundancy). Therefore, we chose PCA since it is the recommended choice as a first step for gait waveform data reduction (Chau, 2001), without loss of temporal information, which generates independent principal components and scores (Deluzio et al., 2014) that were used for the hypothesis tests of this study. The procedure resembles those previously described for analysis of gait-derived waveforms (Brandon et al., 2013; Deluzio and Astephen, 2007; Kirkwood et al., 2011). PCA was performed on 24 gait variables arranged in 24 separate 24×101 data matrices (12 subjects $\times$ 2 sides $\times$ 101 time points per gait cycle). Data related to each measure m were organized in an $n \times p$ matrix X_m. Each row in the matrix X_m represented a temporal series m for each side of each participant. Each column represented the time samples of measure m at one particular instant for each side of all participants.

Each data matrix was mean centered, and the associated covariance matrix was subsequently calculated. The next step in computation involved the eigenvalue decomposition of the covariance matrix; this was achieved according to the principal component model $Z = [U^t X]$, where U is the transformation matrix that rotates the original data observations into a new coordinate system. The columns of U are the eigenvectors of the covariance matrix of the original data set, and are termed principal component (PC) loading vectors (Deluzio and Astephen, 2007). The PCs were extracted in a hierarchical fashion based on the amount of variation they explained; this was calculated by dividing the specific eigenvalue for each corresponding PC by the trace of the covariance matrix (Resende et al., 2016). A criterion of 90% of variance explained was used to determine the number of PCs to retain for data analysis (Resende et al., 2015).

2.4.2. Statistical analysis and interpretation of the PC-scores

The scores of the PCs retained for analysis were tested for normal distribution using Kolmogorov–Smirnov and Shapiro–Wilk tests, and then compared between sides using dependent *t*-tests (for normally distributed scores) and Wilcoxon signed-rank test (for non-normally distributed scores). The significance was set at $\alpha = 0.05$. The effect sizes (e.g. *r-value*) of the comparisons with statistically significant differences were also calculated as follows: if *t*-test was used, $r = \sqrt{\frac{t^2}{t^2 - d_f}}$ where t is the *t*-value and d_f is the degree of freedom; if Wilcoxon signed-rank test was used, $r = \frac{z}{\sqrt{24}}$ where z is the z-score (Field, 2006).

The method of single component reconstruction was used to interpret the differences between sides in PC-scores (Brandon et al., 2013). This method isolates the pattern of variance captured by the specific PC where the sides differed, and had three steps. First, the waveforms representing the operated side and the contralateral side (hereafter referred to as healthy side) pattern of variance on the specific PC were plotted in the same graph (Figs. 2 and 3). The waveforms representing the operated and the healthy sides correspond to a high or low value of the PC-score, depending on which side had higher or lower scores on that specific PC. These waveforms were calculated by first multiplying one standard deviation of the corresponding PC-scores by the PC loading vector and then adding (high) or subtracting (low) the resulting product to the sample mean waveform (Brandon et al.,