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Quadriceps Rate of Force Development affects Gait and Function in People with Knee Osteoarthritis

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Abstract

Objective—Quadriceps weakness exists in people with knee osteoarthritis (OA) but other muscle factors like rate of force development (RFD) may also be affected by knee OA. The purpose of this study was to determine if people with knee OA have deficits in quadriceps RFD, determine if quadriceps RFD would improve predicting knee joint power absorption and generation during free and fast walking, and determine if RFD would improve predicting functional outcomes.

Methods—26 subjects with knee OA and 23 healthy control subjects performed maximal voluntary isometric strength (MVIC) and RFD measures of the quadriceps. Subjects also underwent a 3-D motion analysis of both self-selected free and self-selected fast walking speeds. Joint kinetics were calculated from inverse dynamics.

Results—RFD was not different by group ($p=0.763$), however the OA subjects generated the highest peak RFD at a lower %MVIC ($p=0.008$). Controls walked significantly faster at both free and fast walking speeds ($p=0.001$, $p=0.029$). Knee angles at heel strike and peak knee extension were lower ($p=0.004$, $p=0.027$) in the OA group. During fast walking knee power generation was higher in Controls ($p=0.028$). MVIC and force of highest peak RFD predicted KOOS ADL score in the OA subjects, but only MVIC predicted stair climbing time.

Conclusions—The submaximal force at which peak RFD occurs plays a significant role in knee joint power as well as functional measures in the OA subjects, providing further evidence that factors other than maximal strength are also important in people with knee OA.

Introduction

Knee osteoarthritis (OA) is a condition that affects the entire joint and surrounding muscles, and involves both biological and mechanical factors (Andriacchi, 2009). One of the biological factors that influence mechanics is quadriceps strength, defined by the maximum force generating capacity of the muscles. Quadriceps strength has also been linked to functional deficits and altered movement patterns associated with knee OA (Hurley and Scott 1998; Roos et al. 2011). Not only is quadriceps strength important for daily function in people with knee OA, it has also been linked to later onset and less progression of the disease (Slemenda et al. 1997; Slemenda et al. 1998; Thorstensson et al. 2004). When quadriceps strength is impaired, movement patterns may change that could lead to earlier OA onset or faster progression of the disease. For example, people with knee OA with weak quadriceps use less knee flexion during walking (Childs et al. 2004; Schmitt and Rudolph

2007). Reduced knee flexion during the loading phase of stance may increase the impact, or rate of articular cartilage loading in the knee (Lafortune et al. 1996) and lessen the surface area across which loads are distributed. Higher magnitude or rate of loading when coupled with higher loads due to the co-contraction of opposing knee muscle groups (Kumar et al. 2013) could hasten cartilage degeneration. Despite the evidence that strong quadriceps are beneficial to people with knee OA, greater quadriceps strength has also been associated with an increased likelihood of OA progression in people with knee OA, at least in those with malaligned and lax knees (Sharma et al. 2003). These findings illustrate the complexity of the role of quadriceps strength in people with knee OA and suggest that factors other than strength may be important in maintaining function while minimizing progression of the disease.

One characteristic of people with knee OA is knee joint instability (Schmitt, et al. 2008, Schmitt and Rudolph 2007; Fitzgerald et al. 2004) defined as the sensation of buckling or shifting in the knee. Buckling in the knee is likely to involve tibiofemoral translation and rotation in the transverse plane that would subject the cartilage to shear forces that may hasten its destruction. Historically, the term joint instability has been used synonymously with passive laxity with the presumption that impaired passive restraints would necessarily lead to buckling of the knee. However numerous studies have demonstrated that passive joint laxity and the sensation of instability are not related (van der Esch 2012; Knoop 2012) and that muscle activation may compensate for laxity to reduce instability and improve overall function. For muscles to compensate for excessive joint laxity their activation needs to be timed appropriately and their force must be generated quickly. Therefore, it is plausible that strong quadriceps that can contract rapidly in response to external demands would correlate with better function and lesser joint destruction in people with knee OA; however, this has not been a primary focus of research in this population.

Studies of quadriceps contraction speed are difficult to compare because investigators use varying techniques to study contraction speed. Muscle power is a measure of dynamic strength that is a function of the load on a muscle and the velocity at which the load is moved (McComas 1996). Muscle power has been associated with functional limitations, impaired balance and disability in older adults and has been associated with functional performance more strongly than strength (Orr et al. 2006; Puthoff and Nielsen 2007). Researchers interested in aspects of the speed of muscle force production, like motor unit recruitment and rate coding, often study the rate of isometric force development (RFD) that eliminates shortening speed as a confounding influence on force development (Gordon and Ghez 1987; Van Cutsem et al. 1998). Similar to muscle power, RFD is an important component of physical function as illustrated by Maffiuletti et al., (2010) who found that quadriceps RFD correlates more strongly than maximal quadriceps strength with the activities of daily living score of the Knee Outcome Survey (KOS-ADL) after total knee arthroplasty (Maffiuletti, 2010). Therefore, it is likely that quadriceps RFD is important in function and knee joint stability in people with knee OA, but very few studies have investigated RFD in this population.

Investigators interested in the forces generated by muscles during walking often study joint moments and power calculated using inverse dynamics models (Winter 1990). Inverse dynamics uses the inertial properties of the lower limb segments, actual movements and ground reaction forces to predict the forces and moments necessary to achieve a given gait pattern (Crowninshield et al. 1978). It is important to note that sagittal plane knee joint moments are the net moments which involve the sum of the forces produced by agonist and antagonist muscles that may be active during a given movement. Joint power is the product of the net joint moment and angular velocity (Winter 1990) so joint power may be influenced by the rate of force production that results from agonist and antagonist muscle

activation that are involved in a given movement. During the early stance phase of walking, when the limb is accepting body weight, sagittal plane knee motion is controlled largely by the quadriceps muscles (Perry 1992). People with knee OA have altered sagittal plane knee motions and moments in the early part of the stance phase. It is possible that altered knee motion and moments in the sagittal plane during early stance are the result of impairment in the ability of the quadriceps to produce force rapidly. Since movement and muscle activation patterns influence cartilage loading, and the quadriceps muscles influence movement patterns, it is important to determine the influence of quadriceps net force production and RFD on movement patterns as well as function in people with knee OA. The purpose of this study was to compare net quadriceps RFD in people with knee OA to healthy older adults, determine the association between RFD and physical function, and examine how RFD related to dynamic sagittal knee joint power. We hypothesized that people with knee OA would develop net knee extension force at a slower rate than controls, RFD would improve predicting functional outcomes, and that net quadriceps RFD would explain a unique portion of the variability in sagittal plane knee power during early stance of both free and fast walking as well as the variability in functional measures.

Methods

Subjects

Twenty six subjects with medial knee OA (14 males and 12 females, 65.3 ± 7.8 years old) and 23 healthy age and gender matched control subjects (12 males and 11 females, 63.09 ± 7.66 years old) were recruited for this study. Subjects with valgus knee alignment and symptoms on the lateral side of the knee or under the patella were excluded because of the potential influence on the kinematic and kinetic variables of interest. Subjects were excluded if they had a history of 1. pain or injury in the low back, hips, or knees (other than knee arthritis in the patient population) or ankles; 2. neurological problems; 3. unexplained falls; 4. rheumatoid arthritis; 5. use of assistive device during walking; 6. heart disease or high blood pressure not controlled by medication; or 7. pregnant. Knee OA subjects were recruited from advertisements and from a pool of individuals who had volunteered for previous studies; subjects were screened and if eligible underwent a radiographic evaluation to determine Kellgren and Lawrence (KL) grade (rated by one orthopedic physician) and knee alignment. Subjects with KL Grade I were included if they experienced knee pain on a regular basis, had sought medical attention for the pain and were told by their physician that they had knee OA. All subjects underwent a session of muscle function testing and a separate testing session in which they completed self-report questionnaires of function and 3D motion analysis. In the subjects with bilateral knee OA, the more symptomatic limb was tested. In Control subjects, the limb that was tested was the same limb as the OA subject that they most closely matched by age and sex. All subjects gave informed consent that was approved by the Human Subjects Review Board of the University of Delaware.

Radiographic Assessment

Alignment was assessed using a standing, anterior- posterior radiograph in which the hip, knee, and ankle joints were visible. Alignment was determined by the angle (varus $<180^\circ$, valgus $>180^\circ$) of the mechanical axes of the femur and tibia (Hsu et al. 1990) by one investigator (JDW).

Maximum Voluntary Isometric Contraction Strength (MVIC)

Subjects were seated in a testing chair that was instrumented with a force transducer (SM-250, Interface Inc., Scottsdale, AZ, USA). Subjects were seated with hips flexed ~ 90 degrees and knees flexed ~ 70 degrees and their trunk and thighs were secured with straps to prevent movement. The force transducer was secured to the lower leg approximately 2 cm

above the lateral malleolus with a rigid metal cuff that was covered by a rubber pad and a cloth to minimize material compression at the onset of movement that would have dampened the force signals. Subjects completed 3 maximal voluntary isometric contractions of the quadriceps femoris muscles. Each attempt lasted approximately 3 seconds with at least 1 minute separating each attempt. Subjects were instructed to reach the maximum force as quickly as possible. The highest force generated during the 3 MVIC trials, that had been gravity corrected, was used to normalize the force and RFD outputs generated during testing. Quadriceps strength was defined as the maximum force generated during the 3 MVIC trials.

Rapid Volitional Force Pulses

Subjects were instructed to perform a series of rapid isometric force pulses by kicking “as fast as possible and then relax immediately” after the contraction. Visual targets were not provided because targeting can negatively influence the rate of force production (Gordon and Ghez 1987). Therefore, subjects were instructed to produce “small, medium and large” force pulses in random order until at least 10 pulses at each force level up to the MVIC force were collected in the trial. Each pulse was separated by ~5 seconds (Fig 1). Three trials were collected with at least 1 minute separating each trial. The peak torque for each output was normalized to the torque measured during the MVIC and reported as %MVIC. The first derivative of the force pulses comprised the RFD signal. The peak RFD of each force pulse in all three trials was assessed and the highest peak RFD along with the peak force of the force output associated with the highest peak RFD were recorded and used in the analysis (Fig 1).

Motion Analysis

Subjects performed 10 walking trials each at a self-selected free speed and as “fast as possible while still being safe”. Forty-three retroreflective markers (diameter = 9.5 mm) were placed bilaterally on the pelvis, thighs, lower limbs and feet of the subjects and used to define the limb segments. An 8-camera, 3 dimensional motion capture system (VICON MX, Oxford Metrics, UK) was used to capture movement of the markers at 120 Hz. Two force platforms (Bertec Corp, Columbus, OH) were used to record ground reaction forces at 1080 Hz. Marker trajectories and ground reaction force data were low-pass filtered (Butterworth 4th order, phase lag) at 6 and 50 Hz respectively using Visual 3D (C-Motion, Germantown, MD 20874). Joint angles were calculated from Euler angles (Cole et al. 1993) and joint kinetics, including frontal and sagittal plane knee joint moments and sagittal plane power, were calculated from inverse dynamics, using Visual 3D (C-motion, Germantown, MD). All net joint moments are reported as net external moments. Peak knee joint moments and powers between initial contact and peak knee extension during stance were used in the analysis. Walking speed was calculated from the distance traveled by a marker on the sacral shell during stance using Visual 3D.

Functional Measures

Subjects completed the Knee injury and Osteoarthritis Outcome Score (KOOS), which is self-report questionnaire of pain and function in people with knee problems. The KOOS has five dimensions: pain, symptoms, activities of daily living (ADL), sports and recreation function (Sport), and knee-related quality of life (QOL). All are scored from 0 to 4, and the scores reported as a percentage score (0 = extreme knee problems, 100 = no problems). The KOOS is a valid, reliable, and responsive measure of overall knee joint function in people with OA (Roos et al., 1998). Subjects also performed a timed stair climbing test (SCT) in which they were timed ascending and descending a flight of 12 steps “as fast as you can comfortably and safely” performed by one investigator. The use of 1 hand rail was permitted for safety.

Statistical Analyses

Students t-tests were used to determine group differences in highest peak RFD and force of the highest RFD as well as functional measures. Univariate analyses of variance were used to assess group differences in kinematic and kinetic variables with a fixed factor of Group and Covariate of Speed. This was done because of the strong influence of walking speed on movement patterns (Winter, 1990). Pearson product moment statistic was used to assess the correlations between highest peak RFD and force at which the highest peak RFD occurred and knee joint power absorption and generation during free and fast walking, as well as KOOS-ADL score and SCT. Hierarchical regressions were used to determine if the highest peak RFD or force of the highest RFD contributed to knee power absorption or generation during walking or to functional measures after the influence of MVIC was accounted for. Maximum voluntary isometric contraction force was entered into the regression model first followed by the highest peak RFD and then the force at which the highest peak RFD occurred. As each variable was added to the regression model the unique contribution of that variable to the overall r^2 value was calculated. A statistically significant change in the r^2 value indicates that the variable provides a unique contribution to the overall r^2 after the influence of the previous variables had been accounted for. The contribution of each variable to explaining the variability in the dependent variable was assessed by evaluating whether the change in the r^2 values of each variable was significantly different from zero (Field, 2005). If the change in r^2 values (Δr^2) was statistically different from zero, it was considered to explain a unique portion of the variability in the dependent variable after the previously assessed variables had been accounted for. Statistical significance was set at $p = 0.05$.

Results

Subject characteristics and descriptive statistics are presented in Table 1. The KL score distribution for the medial compartment of the OA subjects was: Grade I: 7; Grade II: 10; Grade III: 10; Grade IV: 0. Knee alignment of the OA subjects ranged from 161 to 181 degrees ($\mu = 173.8$, $\sigma = 4.2$).

Estimated marginal means of kinematic and kinetic variables, adjusted for walking speed, are presented in Table 2. Significant differences between groups in joint kinematics and kinetics during free and fast speed walking were observed in the sagittal and frontal planes. Individuals with knee OA landed with the knee more flexed, reached the same flexion angle as controls during weight acceptance and extended less in the single limb support phase of stance than controls. Sagittal knee moments were no different by group during weight acceptance but knee adduction moments were higher in the OA subjects during free and fast speed walking. No differences were observed in the peak power variables during free speed walking but during fast speed walking peak power generation was lower in the OA subjects and peak power absorption was lower in the OA subjects, although the difference did not reach statistical significance at $p < 0.05$. Correlations between knee power absorption and generation and functional measures with highest peak RFD and Force at Highest peak RFD can be found in Table 3. Hierarchical regression analyses performed on the OA subjects only and the results are shown in Table 4, and Figures 2-4.

Hierarchical Regressions to predict power during free speed walking

The hierarchical regression to predict peak power absorption in early stance during free speed walking (Fig 2) revealed that the variance accounted for (r^2) with the predictor MVIC equaled .000 which was not significantly different from zero ($F_{(1, 24)} = 0.011$, $p = 0.919$). Thus, MVIC did not account for a significant portion of the variance in peak power absorption. Next, Highest Peak RFD was entered into the regression equation resulting in a

change in variance of 0.271 which was statistically different from zero ($\Delta F_{(1, 23)} = 8.536$, $p=0.026$). Thus, Highest Peak RFD accounted for 27% of the variance in peak joint power absorption. The last variable added to the model was Force at Highest Peak RFD that resulted in an R^2 change of 0.156 which was statistically different from zero ($\Delta F_{(1, 22)} = 5.991$, $p=0.006$). This means that the Force at the Highest Peak RFD accounted for 15.6% of the variance in peak joint power absorption during free speed walking after accounting for the effect of the previously entered variables.

The hierarchical regression to predict peak power generation during free speed walking indicated that the variance accounted for (r^2) by the predictor MVIC equaled 0.071 which was not significantly different from zero ($F_{(1, 24)} = 0.120$, $p=0.732$) therefore MVIC alone did not explain a significant portion of the variance in peak power generation during free speed walking. Next, Highest Peak RFD was entered into the model resulting in a change in r^2 of 0.231 which was statistically different from zero ($\Delta F_{(1, 23)} = 6.757$, $p=0.049$). This indicates that Highest Peak RFD accounted for 23.1% of the variance in peak power absorption during free speed walking. The last variable added to the model, Force at Highest Peak RFD, resulted in a change in variance of .067, which was statistically different from zero ($\Delta F_{(1, 22)} = 2.084$, $p=0.047$) and represented nearly 7% of the variance in peak joint power generation during free speed walking after accounting for Highest Peak RFD (Fig 2).

Hierarchical Regressions to predict power during fast speed walking

The hierarchical regression to predict peak power absorption during fast speed walking indicated that the variance accounted for (r^2) by the predictor MVIC equaled 0.444 which was significantly different from zero ($F_{(1, 24)} = 5.886$, $p=0.023$) indicating that MVIC explained 44% of the variance in the peak power absorption during fast speed walking. Next, Highest Peak RFD was entered into the model resulting in a change in variance of 0.078 which was not statistically significant ($\Delta F_{(1, 23)} = 2.473$, $p=0.129$). The last variable added to the model, Force at Highest Peak RFD, resulted in a change in variance of 0.151, which was statistically significant ($\Delta F_{(1, 22)} = 5.797$, $p=0.025$) and represented 15% of the variance in peak joint power absorption during fast walking after accounting for Highest Peak RFD (Fig 3)

The hierarchical regression to predict peak power generation during fast speed walking indicated that the variance accounted for (r^2) by the predictor MVIC equaled 0.492 which was significantly different from zero ($F_{(1, 24)} = 7.685$, $p=0.011$) indicating that MVIC explained 49.2% of the variance in the peak power generation during fast speed walking. Next, Highest Peak RFD was entered into the model resulting in a change in variance of 0.047 which was not statistically different from zero ($\Delta F_{(1, 23)} = 1.506$, $p=0.232$). The last variable added to the model, Force at Highest Peak RFD, resulted in a change in variance of 0.137, which was statistically significant ($\Delta F_{(1, 22)} = 5.239$, $p=0.032$) and represented nearly 14% of the variance in peak joint power generation during fast walking after accounting for Highest Peak RFD (Fig 3)

Hierarchical Regressions to predict Functional Measures

The hierarchical regression to predict KOOS-ADL score (Fig 4) indicated that the variance accounted for (r^2) by the predictor MVIC equaled 0.286 which was significantly different from zero ($F_{(1, 24)} = 9.627$, $p=0.005$) indicating that MVIC accounted for 28.6% of the variance in KOOS-ADL scores. When Highest Peak RFD was added to the model, the change in the r^2 value equaled 0.026 which was not statistically different from zero ($F_{(1, 23)} = 0.883$, $p=0.357$) thus indicating that Highest Peak RFD did not explain any of the variance in KOOS-ADL score. Finally, when Force at Highest Peak RFD was entered into the model, there was a significant change in the r^2 value of 0.197 ($\Delta F_{(1, 22)} = 8.821$,

$p=0.007$) indicating that Force at Highest Peak RFD explained 19.7% of the variance in KOOS-ADL.

The hierarchical regression to predict SCT (Fig 4) indicated that the variance accounted for (r^2) by the predictor MVIC equaled 0.396 which was significantly different from zero ($F_{(1, 24)}=15.732$, $p=0.001$) indicating that MVIC accounted for nearly 40% of the variance in stair climbing times. When Highest Peak RFD and Force at Highest Peak RFD were added to the model, the change in the r^2 value equaled 0.005 and 0.030 respectively, neither of which were statistically different from zero ($F_{(1, 23)}=0.186$, $p=0.670$) and ($\Delta F_{(1, 22)}=1.165$, $p=0.292$), respectively. This indicated that neither of those variables explained any of the variance in SCT in the OA subjects.

Discussion

In the present study the MVIC and peak rate of quadriceps force development were related to biomechanical measures during walking and functional outcomes in people with knee osteoarthritis. The purpose was to determine if muscle performance measures other than maximum force production might influence not only daily function but also influence the subjects' biomechanics of walking that might influence OA progression. We expected people with knee OA to generate force at a slower rate than Control subjects of similar ages and that the RFD would correlate with measures of function and knee power during walking. However, the results did not support these hypotheses. The results showed that while highest peak RFD is no different in people with knee OA than controls, the force of the contraction from which the peak RFD was measured was not only less than the highest percentage of maximum voluntary isometric force output in both groups, it was lower in the subjects with knee OA. In addition, the level of force at which the maximum RFD occurred related to both knee joint power during walking as well as measures of physical function. This supports our speculation that strength (as assessed using the maximum isometric force output) is not the only measure of quadriceps function that relates to biomechanical measures during walking and function in people with knee OA and this finding has implications for treatment of knee OA.

The lack of difference in RFD in people with knee OA and control subjects of similar ages was unexpected, given that studies have shown signs of selective atrophy of fast twitch muscle fibers and evidence of muscle fiber denervation and reinnervation (Berger et al. 2011; Fink et al. 2007). Very few studies have been published to which our results can be compared, however, Maffiuletti et al. (2010) recorded RFD during the first 200 ms of a maximum voluntary isometric contraction in people who had undergone total knee arthroplasty (TKA) 6 months prior to testing. Their results showed that the peak RFD in the TKA limb averaged 898.1 N/s which is consistent with the highest peak RFD generated by our subjects (982 N/sec). The similarity in those results gives us confidence that the methods we used to assess RFD were appropriate.

The biomechanical variables during walking that were demonstrated by the subjects in the present study are similar to those reported by others. For example, on average, the subjects in our study landed with the knee flexed 9° , continued flexing to 21° then extended to approximately 10° of knee flexion as they entered single limb support. These numbers are consistent with those reported by Heiden et al., (2009) who observed that OA subjects landed in approximately 8° of flexion, flexed to $\sim 19^\circ$ then extended to $\sim 9^\circ$ of flexion during self-selected free speed walking. Similarly, the external knee flexion moments observed in our subjects averaged (OA: 0.45 Nm/kg; Controls: 0.43 Nm/kg) which were similar to the data reported by Baert et al. (2012), who reported peak external knee flexion moments of approximately 0.4 Nm/kg in OA and control subjects. Since the results from our

subjects are similar to those reported by others, we have confidence that our subjects walked in a way that has been demonstrated in other studies of people with knee OA and were not unusual in terms of physical behavior.

Although the biomechanical variables of our subjects appear typical of people with knee OA, they rated themselves higher on the functional outcome scale than the subjects in other studies. The KOOS-ADL scores of the subjects in the present were KOOS-ADL: = 72, whereas those reported by Baert et al. (2012) were KOOS-ADL: =66) and Heiden et al. (2009) (KOOS-ADL: =60). The higher function in the subjects in the present study may have occurred because we included subjects with KL grade I who reported experiencing knee pain for which they had sought the care of a physician but whose radiographs had doubtful joint space narrowing or only possible osteophytic lipping. We chose to include these subjects because although the KL grade indicated very little radiographic evidence of joint damage (KL grade I), they had enough pain or lost function to seek care of a physician. Most studies of people with knee OA include subjects only if they have KL grade II or higher however, we chose to include individuals who had pain and functional limitations that were substantial enough to seek medical care. Therefore, different results may have been observed if only people with more severe OA were included.

Another unexpected finding is that the Highest Peak RFD was not achieved at the highest force output, which might have been expected based on previous reports in the literature. Previous studies have demonstrated that the time to reach peak force across multiple force levels is relatively invariant (Bellumori et al. 2011; Freund and Budingen 1978). Corcos et al. (1989) described this phenomenon as a speed control or speed-sensitive control strategy that has been observed in young healthy subjects (Corcos et al. 1989; Freund and Budingen 1978). If time to peak force remains constant across force levels, then it follows that RFD will increase in a strong linear fashion. If that were the case, the fastest RFD would occur in conjunction with the largest force output. However, in the present study, the force at which the highest peak RFD occurred was at forces that were less than the highest voluntary isometric contraction force achieved in the groups. , It is plausible that this finding is evidence of altered motor control strategies in these subjects, but further investigation is needed to verify this speculation.

To illustrate our interpretation of these findings, RFD and time to peak force from each force pulse generated by one randomly chosen subject are shown in Figure 5. For pulses in which the subject generated force between 15-50% MVIC, the time to reach peak force was relatively invariant and equaled ~100 ms. This resulted in a linear rise in RFD from 15 to 50% MVIC. Beyond 50% MVIC, as the force magnitude increased, the time it took to reach peak force increased. In fact, the highest force produced by the subject was ~85% MVIC and the time to reach peak force was ~275 ms. The dashed circle represents the contraction with the fastest RFD that occurred when the subject had generated about 53% MVIC. Had the time to reach peak force remained invariant across all force levels, the highest RFD would have been associated with the highest force contraction, but that was not the case. The force at which the highest RFD was found in the OA subjects was 64% MVIC, which was significantly lower than the Controls (71% MVIC) and suggests that impairments in rapid force generation are more profound in older adults with OA than without OA.

Wierzbicka et al. (1991) also found an increase in the time to peak force with force magnitude, but in people with Parkinson's Disease (Wierzbicka et al. 1991). The authors attributed the findings to deficits in single motor unit recruitment in the people with Parkinson's Disease who lacked doublet activations (i.e. two consecutive motor unit action potentials occurring less than 20 ms apart (Burke et al. 1970)) in motor units as they were recruited. Reduction in firing rates during high force contractions was reported by Leong, et

al., (1999) in health older adults. Leong et al., (1999) found that motor unit firing rates during submaximal contractions in untrained older adults were no different from those used by older adults who had undergone strength training. However, at 100% of the MVIC, those subjects who had not undergone the strength training had lower firing rates than the strength trained counterparts. It is possible that in our subjects, the firing rates of muscles at forces below 50% MVIC could be modulated to maintain the same time to reach peak force regardless of force magnitude, but the same was not true for higher force magnitudes. While this speculation is plausible, other factors such as stiffness in the muscles and non-contractile tissues as well as pain may play a role in RFD in this population so further study of motor unit recruitment as well as muscle-tendon stiffness would provide greater insight into the mechanisms underlying our findings.

While no differences were observed in the Highest Peak RFD between the OA and control subjects, the correlations between Highest Peak RFD and Force at the Highest Peak RFD with knee power absorption and generation during free speed walking were statistically significant and moderately high in the OA subjects. Hierarchical regression revealed that while MVIC did not explain a significant portion of the variance in the knee power variables during free speed walking, Highest Peak RFD and Force at the Highest Peak RFD explained unique portions of the variance in the predicted variables. These findings support our speculation that factors other than MVIC are involved in some aspects of function in this population. It is curious that Highest Peak RFD did not explain any of the variance in the knee power at the fast walking speed; instead, it was the MVIC and Force at the Highest Peak RFD that explained unique proportions of the variance in the fast walking knee power absorption and generation. The fact that MVIC was the strongest predictor of SCT is not surprising given that stair climbing would require more knee extensor force than walking or other activities of daily living that are less challenging. It appears that aspects of force output relating to speed may also significantly influence the muscle's ability to control the knee during specific dynamic tasks rather than all dynamic tasks. Mechanical joint powers represent the functional role of anatomical structures as they shorten and lengthen under tension (Eng & Winter 1995). According to the results of this study, during walking, the speed of the contraction and the force that can be produced rapidly play a significant role in controlling knee flexion (power absorption) and knee extension (power generation) during the loading phase of gait. The ability to flex and extend the knee may be vital in this population, lessening the "stiffened gait", therefore, the relationship between RFD and knee joint powers warrants further investigation.

There are several reasons why increased RFD at higher levels of force may be of clinical importance to people with knee OA. Fall prevention and knee stabilization may both require sufficient force delivered at sufficiently fast rates, between 0-200ms, (Suetta et al. 2004) and our results show an increased time to peak force at higher force outputs, indicating that if higher levels of force are needed they may not be generated within the necessary time constraints. This may be the result in deficits in quadriceps function or due to excessive muscle co-contraction. Analysis of muscle activation patterns during the RFD and walking trials is currently underway and will be reported in a future paper. In recent years much attention has been paid to the influence of age related changes in neuromuscular performance on risk of falls in older adults (Caserotti et al. 2008; Foldvari et al. 2000). Knee extensor muscle power, which is a function of both the magnitude of force and speed of muscle shortening, relates to functional capacity in older adults and power training is often recommended in exercise regimens for elderly individuals (Marsh et al. 2009; Muehlbauer et al. 2012; Paterson et al. 2007). Knee osteoarthritis typically develops with age and falls in people with knee OA are not uncommon (Brand et al. 2005), so it is plausible that improving the magnitude of force that can be generated rapidly may help reduce falls in people with knee OA. Though this study did not measure muscle power, the reduction in

force associated with the peak RFD may be influenced by similar mechanisms since power is a product of both force and velocity but further evidence is needed to support hypothesis.

In addition to its influence on risk of falls and on functional activities, it is possible that the ability to rapidly generate higher levels of force could affect the progression of knee OA. Recently, studies have demonstrated that the velocity of movement between the tibia and femur may contribute to an increase in “plowing” that creates mechanical disruption of the articular surfaces (Anderst and Tashman 2009). Whether or not muscles that are capable of responding to and generating sufficient levels of force quickly can stabilize and control movement between joint segments, mitigating frictional forces, is beyond the scope of this study. Based on previous findings that greater quadriceps strength was associated with an increased likelihood of OA progression in people with malaligned and lax knees (Sharma et al. 2003) it appears that factors other than maximal strength may be involved. Whether the rate of quadriceps force development or the force of highest peak RFD influences factors that lead to joint degeneration remains to be seen but the findings of this study provide support for our speculation that studying quadriceps performance beyond deficits in maximum force generating capacity of the muscle may provide valuable insights into motor control deficits that could be addressed through exercise to improve function and lessen OA progression.

Several limitations exist in the present study. First, power that is calculated using biomechanical data from inverse dynamics models relies on the use of net joint moments that represent the net moment that can result from more than one combination of agonist and antagonist muscle activations. Therefore, it is possible that greater hamstring activation during early stance could influence the kinetic variables and that the time to peak force data from quadriceps contractions alone is less important. Second, although all subjects in the OA group had knee OA based on their symptoms (they had sought out the care of a physician for knee pain and had been told they had knee OA) some of the subjects' radiographs had been rated at KL grade 1, which indicates minimal radiographic changes in the structures of the knee joint. Therefore, it is possible that the underlying disease process of some of the subjects was so mild that differences in muscle performance and changes in the biomechanical variables used during walking had not yet developed.

The maximum voluntary isometric contraction forces were normalized to BMI, which is a measure that include body mass and height squared. We did this in order to account for the effect of height, and therefore the length of the tibia that produced the moment of force generated during the MVIC tests as well as body mass, which could influence the force generated by the quadriceps that are accustomed to controlling more or less body mass during daily activities. However, the same BMI may result from people with different combinations of mass and height so our results may have been different had we not normalized the MVIC data in this manner.

The number of subjects included in this study was relatively small, so it is possible that the statistical power was too low to find differences by group. Another limitation to the study was using walking speed as a covariate for the gait analysis. Walking speed is slower in people with knee OA, and walking speed influences joint angles, moments and power generated during walking. Thus we used speed as a covariate to ensure that differences we observed in the walking variables were due to differences in the subjects rather than differences in walking speed. None-the-less, using speed as a covariate may have removed portions of the main effect or group from the analysis (Wilson, 2012). Finally, when examining movement compensation strategies it is important to examine all the joints involved, such as the ankle, knee and hip during walking. The focus of this study was not a comprehensive analysis of movement strategies but rather was based on a priori hypotheses

related to the rate of force production on walking variables and knee function. Future analyses of walking patterns involving other joints may reveal additional insights into ways that muscle RFD movement patterns throughout the lower limbs and trunk as well.

In this study, we demonstrated that the highest rate of quadriceps force development was no different in the OA and control subjects, however, the force at which the highest peak RFD occurred was lower in the OA subjects. Moreover, the relationships between the highest peak RFD and the force at which the highest peak RFD occurred and biomechanical gait variables including peak power absorption and generation appeared in the OA subjects only. This demonstrates that rate of force production influences walking variables that have been shown previously to relate to function, particularly in older adults. Further investigation is warranted to investigate whether emphasizing rapid muscle contractions along with strengthening activities in exercise programs will help people with knee OA maintain function and slow disease progression.

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Fig 1. Force magnitude (top) and rate of force development (bottom) of force pulses of various magnitudes

Fig 2. Hierarchical regression predicting self-selected walking speed: peak power absorption (top) and peak power generation (bottom) by entering into the model MVIC first, followed by Highest Peak RFD then Force at Highest Peak RFD

Fig 3. Hierarchical regression predicting fast walking speed: peak power generation (top) and peak power generation (bottom) by entering into the model MVIC first, followed by Highest Peak RFD then Force at Highest Peak RFD

Fig 4. Hierarchical regression predicting self-selected walking speed: KOOS-ADL score (top) and stair climbing time (bottom) by entering into the model MVIC first, followed by Highest Peak RFD then Force at Highest Peak RFD

Fig 5. Rate of Force development (top) and time to peak force (bottom) from one subject with knee OA. Each point represents the data from one force pulse.

Table 1
Subject characteristics

	OA ()	Control ()	p-value
Age (yrs)	65.30 (7.80)	63.10 (7.70)	0.705
Body Mass (kg)	93.3 (16.4)	75.6 (15.7)	0.000*
BMI	30.60 (4.60)	25.80 (3.80)	0.000*
MVIC (N)	444.50 (165.70)	486.10 (150.70)	0.365
Highest Peak RFD (N/sec)	982.90 (200.90)	999.60 (183.60)	0.763
Force at Highest Peak RFD (%MVIC)	64.00 (9.0)	71.10 (9.00)	0.008*
Free Walking Speed (m/s)	1.27 (0.04)	1.46 (0.03)	0.001*
Fast Walking Speed (m/s)	1.84 (0.10)	2.05 (0.10)	0.029*
KOOS-ADL Score	72.30 (19.00)	99.70 (0.91)	0.000*
Stair Climbing Time (s)	13.08 (5.06)	8.82 (1.82)	0.000*

BMI = Body Mass Index

MVIC = Maximum Voluntary Isometric Contraction

RFD = Rate of Force Development

KOOS-ADL = Knee injury Osteoarthritis Outcome Scale - Activities of Daily Living

* Statistically significant at p 0.05

Table 2
Knee Kinematic and Kinetic variables adjusted for walking speed

	Variable	OA ()	Control ()	p-value
Free Speed	Flexion - IC (deg)	-9.40 (0.80)	-5.80 (0.80)	0.004*
	Pk Flexion - LR (deg)	-20.90 (1.10)	-19.90 (1.20)	0.525
	Pk Extension - MSt (deg)	-10.80 (1.10)	-6.90 (1.20)	0.027*
	Pk Flexion Moment - LR (Nm/kg)	0.45 (0.03)	0.42 (0.04)	0.555
	Pk Adduction Moment - LR (Nm/kg)	-0.48 (0.03)	-0.32 (0.03)	0.001*
	Pk Power Absorption - LR (W/kg)	-0.55 (0.06)	-0.69 (0.06)	0.123
	Pk Power Generation - LR (W/kg)	0.29 (0.045)	0.36 (0.04)	0.193
Fast Speed	Flexion - IC (deg)	-11.90 (0.70)	-8.60 (0.80)	0.005*
	Pk Flexion - LR (deg)	-24.70 (1.20)	-22.60 (1.20)	0.242
	Pk Extension - MSt (deg)	-10.67 (1.01)	-3.90 (1.10)	0.000*
	Pk Flexion Moment - LR (Nm/kg)	0.62 (0.06)	0.57 (0.06)	0.549
	Pk Adduction Moment - LR (Nm/kg)	-0.61 (0.04)	-0.44 (0.04)	0.003*
	Pk Power Absorption - LR (W/kg)	-1.15 (0.13)	-1.48 (0.13)	0.089
	Pk Power Generation - LR (W/kg)	0.60 (0.07)	0.86 (0.08)	0.028*

Negative angles indicate joint flexion; moments are internal moments so negative frontal plane moments indicate net internal adduction moments

S.E.=Standard Error

Pk=peak

IC = Initial Contact

LR=Loading Response

MSt=Midstance

* Statistically significant at p = 0.05

Table 3
Correlations between highest peak RFD, Force at Highest peak RFD with kinetic and functional variables

		Peak Power Absorption Free Speed	PK Power Generation Free Speed	PK Power Absorption Fast Speed	PK Power Generation Fast Speed	KOOS ADL	SCT
Knee OA	Highest PKRFD	Pearson Correlation Sig. (2-tailed)	-.520** .006	.476* .014	-.281 .165	.164 .423	.068 .743
	Force at Highest PKRFD	Pearson Correlation Sig. (2-tailed)	-.573** .002	.424* .031	-.398* .044	.383 .054	-.028 .892
	Highest PKRFD	Pearson Correlation Sig. (2-tailed)	-.198 .366	.008 .971	-.231 .290	-.103 .641	-.208 .341
Controls	Force at Highest PKRFD	Pearson Correlation Sig. (2-tailed)	-.279 .197	.273 .208	-.392 .064	-.166 .450	-.060 .786

** Correlation is significant at the 0.01 level (2-tailed)

* Correlation is significant at the 0.05 level (2-tailed)

Table 4
Results of hierarchical regressions to predict peak power during walking and functional measures

Condition	Dependent Variable	Variables entered into the model in each step	R	Model R ²	ΔR^2	p value	df	β
Free Speed Walking	Peak Power Absorption	1. MVIC	.021	.000	.011	.919	1, 24	-.088
		2. Highest Peak RFD	.521	.271	8.536	.026*	1, 23	-.328
		3. Force @ Highest Peak RFD	.653	.427	5.991	.006*	1, 22	-.445
Fast Walking	Peak Power Generation	1. MVIC	.071	.005	.120	.732	1, 24	.114
		2. Highest Peak RFD	.481	.231	6.757	.049*	1, 23	.350
		3. Force @ Highest Peak RFD	.545	.297	2.084	.047*	1, 22	.290
	Peak Power Absorption	1. MVIC	.444	.197	5.886	.023*	1, 24	-.510
		2. Highest Peak RFD	.524	.275	2.473	.129	1, 23	-.090
		3. Force @ Highest Peak RFD	.653	.348	5.797	.025*	1, 22	-.438
Functional Measures	Peak Power Generation	1. MVIC	.492	.243	7.685	.011*	1, 24	.556
		2. Highest Peak RFD	.538	.289	1.506	.232	1, 23	.036
		3. Force @ Highest Peak RFD	.653	.426	5.239	.032*	1, 22	.416
	KOOS-ADL	1. MVIC	.535	.286	9.627	.005*	1, 24	.612
		2. Highest Peak RFD	.559	.313	.883	.357	1, 23	-.054
		3. Force @ Highest Peak RFD	.714	.509	8.821	.007*	1, 22	.499
	Stair Climbing Time	1. MVIC	.629	.396	15.732	.001*	1, 24	-.660
		2. Highest Peak RFD	.633	.401	.186	.670	1, 23	.154
		3. Force @ Highest Peak RFD	.656	.431	1.165	.292	1, 22	-.195

* p<0.05 indicating that a significant change in the r² value to predict the relevant dependent variable