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1 Title: Quantifying varus thrust in knee osteoarthritis using wearable inertial sensors: a proof of concept

2

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21

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23 **Abstract**

24 *Background:* Varus thrust during walking, visualized as excessive frontal plane knee motion during
25 weight acceptance, is a modifiable risk factor for progression of knee osteoarthritis. However, visual
26 assessment does not capture thrust severity and quantification with optical motion capture is often not
27 feasible. Inertial sensors may provide a convenient alternative to optical motion capture. This proof-of-
28 concept study sought to compare wearable inertial sensors to optical motion capture for the
29 quantification of varus thrust.

30 *Methods:* Twenty-six participants with medial knee osteoarthritis underwent gait analysis at self-
31 selected and fast speeds. Linear regression with generalized estimating equations assessed associations
32 between peak knee adduction velocity or knee adduction excursion from optical motion capture and
33 peak thigh or shank adduction velocity from two inertial sensors on the lower limb. Relationships
34 between inertial measures and peak external knee adduction moment were assessed as a secondary
35 aim.

36 *Findings:* Both thigh and shank inertial sensor measures were associated with the optical motion capture
37 measures for both speeds ($P < 0.001$ to $P = 0.020$), with the thigh measures having less variability than
38 the shank. After accounting for age, sex, body mass index, radiographic severity, and limb alignment,
39 thigh adduction velocity was also associated with knee adduction moment at both speeds (both $P <$
40 0.001).

41 *Interpretation:* An inertial sensor placed on the mid-thigh can quantify varus thrust in people with
42 medial knee osteoarthritis without the need for optical motion capture. This single sensor may be
43 useful for risk screening or evaluating the effects of interventions in large samples.

44 **Keywords:** angular velocity, gyroscope, motion capture, adduction

45 **1. Introduction**

46 Knee osteoarthritis (OA) is a leading cause of disability among older adults¹ and most commonly affects
47 the medial tibiofemoral compartment of the knee joint². While age, sex, genetics, and other non-
48 modifiable factors have been implicated in OA pathogenesis, gait patterns leading to increased or
49 abnormal biomechanical joint loading also play a role and are frequently targeted in interventions³. A
50 common gait abnormality in people with medial knee OA is varus thrust, an excessive ‘bowing-out’ knee
51 motion in the frontal-plane during ambulation as the limb accepts weight with a return towards a more
52 neutral alignment in late stance and swing^{4,5}. Varus thrust has been reported to be present in 12% to
53 46% of individuals with medial knee OA and has been associated with radiographic disease severity⁶ and
54 progression⁵. Cross-sectionally, those with varus thrust have a five-times greater odds for higher pain
55 during walking and standing than those without it⁷. Individuals with knee OA who exhibit varus thrust
56 also exhibit greater peak external knee adduction moments (EKAM) during gait⁵, an indication of medial
57 tibiofemoral load⁸ which has been reported to be a risk factor for future OA progression⁹. Thus,
58 interventions to reduce varus thrust may lead to reduced pain and slow structural worsening in
59 individuals with medial compartment knee OA.

60 To aid in the development of effective interventions, it is important to accurately and reliably
61 identify the presence of varus thrust. Typically, varus thrust is assessed through a subjective, visual
62 evaluation of walking^{5-7,10-14}. While these assessments are used clinically, they only provide a
63 dichotomous categorization (present/absent) without any indication of severity. To overcome this
64 limitation, optical motion capture has been used to objectively quantify biomechanical parameters as
65 surrogate measures of varus thrust^{5,6,15-21}, including knee adduction velocity⁶ and knee adduction
66 angular excursion²¹. However, while optical motion capture provides detailed information on joint
67 kinematics and kinetics, these systems require expensive equipment, time-consuming data collections
68 run by skilled technicians, and a large calibrated measurement volume, making their clinical use

69 infeasible. Additionally, analyses conducted in a laboratory environment do not always reflect typical
70 walking in real-world settings²². In contrast, small, low-cost wearable inertial sensors have become
71 increasingly popular for collecting biomechanical data in free-living conditions and may provide a
72 convenient alternative to optical motion capture systems for quantifying varus thrust²³.

73 The primary aim of this study was to compare data from a single wearable inertial sensor to
74 surrogate measures of varus thrust captured using optical motion capture technology during self-
75 selected and fast speed walking in individuals with medial compartment knee osteoarthritis. We
76 hypothesized that measures of frontal plane segment velocity from single inertial sensors placed on the
77 thigh or shank would be significantly associated with measures from optical motion capture based on
78 previously reported agreement between inertial sensor and optical motion capture kinematic and
79 kinetic measures^{24,25}. For a secondary aim, we hypothesized that the inertial sensor measures would be
80 associated with EKAM after adjusting for confounders.

81

82 **2. Methods**

83 *2.1. Participants*

84 Participants were recruited using advertisements online and in local newspapers from October 2017 to
85 May 2019. Inclusion criteria were age between 45-80 years, body mass index (BMI) ≤ 40 kg/m², and at
86 least one knee meeting the American College of Rheumatology clinical or radiographic criteria for knee
87 OA²⁶ with primarily medial tibiofemoral compartment involvement (medial joint space narrowing
88 identified from weight bearing knee radiographs). Exclusion criteria were regular use of a walking aid,
89 inflammatory arthritis, lower limb total joint replacement, neurological conditions, muscular disease, or
90 other conditions/treatments affecting gait. This study was approved by the Boston University
91 Institutional Review Board, Boston, USA, and all individuals provided written informed consent prior to
92 radiographic screening and data collection.

93

94 *2.2. Radiographs and Assessment of Symptoms*

95 All participants underwent three radiographs: (1) bilateral weight-bearing posterior-anterior flexed knee
96 radiograph using a Synaflexer positioning frame (BioClinica, Princeton, NJ, USA) for assessment of
97 Kellgren-Lawrence grade (KLG) and assessment of medial and lateral tibiofemoral compartment
98 involvement based on the Osteoarthritis Research Society International (OARSI) atlas, (2) bilateral knee
99 sunrise view for assessment of patellofemoral involvement, and (3) bilateral standing long limb view for
100 measurement of static mechanical axis alignment. An experienced clinician assessed OA severity (KLG)
101 and determined the most involved compartment (medial or lateral) for each knee. Inter-reader
102 reliability for KLG of 0.79 has been reported previously²⁷. Static alignment was calculated using OsiriX
103 open-source software (www.osirixviewer.com)²⁸ as the angle formed by the intersection of the line
104 between the center of the femoral head and midpoint of the femoral epicondyles, and the line between
105 the midpoint of the femoral epicondyles and midpoint of the malleoli. All participants also completed
106 the Knee Injury and Osteoarthritis Outcome Score (KOOS)²⁹. All knees with KLG < 2 were excluded from
107 the analyses.

108

109 *2.3. Optical motion capture*

110 Ground reaction force (GRF) and kinematic data were collected from both legs for all participants while
111 walking along a 15-meter walkway at self-selected and fast speeds. A passive optical motion capture
112 system with 12 infrared cameras and one video camera (Qualisys Medical, Gothenburg, Sweden) was
113 used to capture kinematic data at 250 Hz. Twenty-six spherical retroreflective markers were attached to
114 bony landmarks of the trunk (manubrium, C7 spinous process, T8 spinous process, and the right and left
115 acromion), pelvis (right and left anterior superior iliac spines, superior aspects of iliac crests, right and
116 left posterior superior iliac spines, and sacrum), and lower extremities (greater trochanters, lateral and

117 medial femoral epicondyles, lateral and medial malleoli, first and fifth metatarsal heads) which were
118 used to identify joint centers during a static standing trial (Figure 1A). Rigid clusters of four markers
119 were placed on the shank and thigh segments, and markers were placed on the posterior aspect of the
120 heel, medial and lateral mid-foot, 2nd metatarsal head, and lateral aspect of 5th metatarsal head for
121 segment tracking (Figure 1A). Three force platforms (AMTI, Watertown, MA, USA) were used to collect
122 GRF data at 2000 Hz synchronously with the motion data. Walking speed was measured using a timing
123 system (Brower Timing Systems, Draper, UT, USA). Self-selected and fast walking target speeds were
124 determined after 3-4 practice trials. For self-selected speed trials, participants were instructed to “walk
125 across the room towards the door with a purposeful pace.” For fast walking trials, participants were
126 instructed to “walk as quickly as you comfortably can without running and without going faster than
127 what you feel is safe for you.” Any trials that were outside $\pm 5\%$ of the target walking speeds were
128 excluded. Four to six clean force plate foot strikes were collected for each foot. Each participant wore
129 laboratory-provided shoes (Gel-Cumulus 19, ASICS, Kobe, Japan).

130 Within the motion capture software, marker trajectories were identified and any gaps in data
131 were filled using polynomial splines for gaps of less than 10 frames and trajectory matching for larger
132 gaps. All data were then imported into Visual3D (C-Motion, Germantown, MD, USA). Marker and GRF
133 data were filtered using low-pass, 4th order Butterworth filters with cut-off frequencies of 6 Hz and 12
134 Hz, respectively, as the majority of movement during walking occurs below these frequencies. Joint
135 kinematics were calculated from the marker data using Euler angles (x-y-z) and right-handed coordinate
136 systems. Initial and final contact for each clean foot strike were identified from the GRF using a
137 threshold of 20 N to define stance phase.

138 The primary measure from the optical motion capture system was peak knee adduction velocity
139 and the secondary measure was knee adduction excursion. Chang et al. suggested peak knee adduction
140 velocity as an appropriate biomechanical index to quantify varus thrust as it closely corresponded with

141 visual assessments and captures both the direction and speed of movement⁶. As the majority of the
142 varus thrust movement occurs in early stance⁶, peak knee adduction velocity during the first half of
143 stance phase was recorded (Figure 2B). It should be noted that because the joint angle (Euler angle) is a
144 vector quantity, it is not possible to compute the joint angular velocity by taking the first derivative of
145 the joint angle³⁰. We used X-Y-Z cardan sequence to calculate the knee joint angle (shank relative to
146 thigh); the knee joint angular velocity was calculated as the angular velocity of the shank relative to the
147 thigh^{30,31}. The secondary motion capture varus thrust measure, knee adduction excursion, was
148 calculated as the difference between the knee adduction angle at initial contact and the maximum knee
149 adduction angle during the first half of stance⁶ (Figure 2C). Inverse dynamics were used to calculate the
150 EKAM, normalized to body weight and height (% bodyweight-height), and the peak EKAM was extracted
151 from the first half of stance (Figure 2D). All variables were averaged across all clean foot strikes for each
152 leg.

153

154 *2.4. Inertial motion capture*

155 Small, lightweight inertial sensors (Trigno™ IM Sensor, Delsys, Inc., Natick, MA, USA) were used
156 concurrently with the optical motion capture system. Each sensor measured 37mm x 26mm x 15mm,
157 weighed 14.7g, and consisted of a triaxial accelerometer ($\pm 16g$), a triaxial gyroscope (± 2000 degree/s),
158 and a triaxial magnetometer ($\pm 1000\mu T$). Initially, three sensor locations were tested for each limb
159 (Figure 1B): 1) lateral mid-thigh (attached to the thigh segment optical motion capture marker cluster),
160 2) lateral mid-shank (attached to the shank segment optical motion capture marker cluster), and 3)
161 lateral distal shank (attached directly to the skin on the lateral aspect of the distal tibia proximal to the
162 lateral malleolus). The distal shank placement was included with the assumption that it would be more
163 convenient for use in a non-laboratory environment. However, interim analyses showed poor
164 association between optical motion capture data and data from the distal sensor, particularly during fast

165 speed walking, and this sensor was perceived as uncomfortable by a few participants. For these reasons,
166 use of the distal sensor placement was discontinued for the final 11 participants (22 knees) enrolled in
167 this study and the data from this sensor are not presented here. For all sensor placements, while a
168 general orientation of the sensors was specified (i.e. arrow side 'up'), the placement along the length of
169 the leg and anterior-posterior position were not constrained, nor were any calibration procedures
170 performed, in order to better replicate placement of the sensors in a clinical setting and/or by untrained
171 individuals (e.g. participants).

172 Data were recorded from each sensor at 148 Hz, upsampled to 2000 Hz, and time synchronized
173 with the optical motion capture system. The frontal plane component of the raw gyroscope data was
174 used as a measure of segment adduction velocity. The gyroscope component was chosen as it captures
175 angular velocity as compared to accelerometer components which capture linear acceleration. Peak
176 segment adduction velocity in degrees per second, which was calculated as the peak value between
177 initial contact and midstance, was extracted for each trial for each sensor (Figure 2A) and averaged
178 across trials for each leg. Outcomes used in the analyses included peak thigh adduction velocity from the
179 mid-thigh sensor (mid-thigh adduction velocity) and peak shank adduction velocity from the mid-shank
180 sensor (mid-shank adduction velocity) for each walking speed. It should be noted that the thigh
181 adduction velocity is recorded as positive and shank adduction velocity is recorded as negative (Figure 2,
182 Table 2) given the nature of segmental motion and orientation of the sensor coordinate systems (Figure
183 1B).

184

185 *2.5. Statistical Analyses*

186 For the primary aim, univariate regression models with generalized estimating equations (GEE) were
187 used to assess the relationships between the inertial sensor measures and the optical motion capture
188 measures, separately for self-selected and fast speed walking. The GEEs allowed us to account the

189 correlation between knees within each person. For the secondary aim, multivariate regression models
190 with GEE were used to assess the relationships between EKAM and the inertial sensor measures,
191 separately for each speed, while adjusting for a number of confounders (specifically age, sex, BMI, KLG,
192 and static alignment) that may affect both varus thrust¹³ and EKAM³²⁻³⁴. While pain can also affect
193 EKAM³², we hypothesized that it acts as a mediator rather than a confounder on the causal pathway (i.e.
194 varus thrust causes pain rather than the other way around), and thus did not include it in the models. All
195 models were constructed including a term for leg and an interaction term between exposure and leg
196 to determine whether the association of the exposure and the outcome differed by leg. If there was
197 no interaction between exposure and leg (i.e. this interaction term was not significant), the model was
198 re-run without the exposure by leg interaction term. All analyses were performed using IBM SPSS
199 (Armonk, NY, USA) in all knees with KLG ≥ 2 . Significance was set at $\alpha = 0.05$.

200

201 **3. Results**

202 One hundred and sixty-three individuals underwent telephone screening, 82 passed the initial screening
203 process, 59 underwent a radiographic screening visit, and 26 individuals (16 female) were deemed
204 eligible for this study (Figure 3, Table 1). The number of knees across analyses differed depending on
205 useable data available for each inertial sensor (Figure 3). Average values for each inertial and optical
206 motion capture measure are reported in Table 2.

207 Both mid-thigh and mid-shank adduction velocity were associated with knee adduction velocity
208 and excursion during self-selected and fast speed walking. An increase of $10.0^\circ/\text{s}$ in mid-thigh adduction
209 velocity was associated with an increase in knee adduction velocity of $6.1^\circ/\text{s}$ during self-selected speed
210 walking ($P < 0.001$, Figure 4A) and $5.3^\circ/\text{s}$ during fast walking ($P = 0.001$, Figure 4B), and an increase in
211 knee adduction excursion of 0.35° during self-selected speed walking ($P = 0.005$, Figure 4C) and 0.33°
212 during fast walking ($P = 0.020$, Figure 4D). Similarly, an increase of $10.0^\circ/\text{s}$ in mid-shank adduction

213 velocity was associated with an increase in knee adduction velocity of $3.4^{\circ}/s$ during self-selected speed
214 walking ($P < 0.001$, Figure 4A) and $2.2^{\circ}/s$ during fast walking ($P = 0.005$, Figure 4B), and an increase in
215 knee adduction excursion of 0.20° during both self-selected ($P = 0.004$, Figure 4C) and fast ($P < 0.001$,
216 Figure 4D) speed walking.

217 After accounting for age, sex, BMI, KLG, and static alignment, an increase of $10.0^{\circ}/s$ in mid-thigh
218 adduction velocity was associated with an increase in EKAM of 0.16 % bodyweight-height ($P < 0.001$)
219 during self-selected speed walking and an increase of 0.17 % bodyweight-height ($P < 0.001$) during fast
220 speed walking. For the models investigating the relationship between mid-shank adduction velocity and
221 EKAM during self-selected and fast speed walking, the interaction term between leg and mid-shank
222 adduction velocity was significant, thus separate models were run for left and right legs. For left legs ($n =$
223 18), an increase in $10.0^{\circ}/s$ in mid-shank adduction velocity was associated with an increase in EKAM of
224 0.10 % bodyweight-height ($P = 0.010$) during self-selected speed walking and an increase in 0.17 %
225 bodyweight-height ($P < 0.001$) during fast speed walking, after accounting for confounders. For right legs
226 ($n = 23$), mid-shank adduction velocity was not associated with EKAM at either self-selected speed ($P =$
227 0.88) or fast speed ($P = 0.21$).

228

229 **4. Discussion**

230 This proof-of-concept study showed that the measures from single inertial sensors were associated with
231 surrogate measures of varus thrust obtained using optical motion capture. Furthermore, supporting our
232 secondary hypothesis, mid-thigh adduction velocity was significantly associated with peak EKAM after
233 adjusting for confounders. These results suggest that inertial sensors should be further investigated as a
234 tool to objectively quantify varus thrust in clinical settings where optical motion capture is not feasible
235 and visual assessment is insufficient. The ability to quickly and accurately quantify varus thrust in clinical

236 or other real-world settings could lead to better identification and treatment of those at risk of OA
237 progression due to varus thrust.

238 Both inertial sensor metrics – mid-thigh adduction velocity and mid-shank adduction velocity –
239 were associated with both optical motion capture thrust measures at both walking speeds. The mid-
240 shank data, however, had greater variability than the mid-thigh data, e.g. an adduction velocity range of
241 $165^{\circ}/s$ for the mid-shank versus $99^{\circ}/s$ for the mid-thigh during self-selected speed walking (Figure 4,
242 Table 2), suggesting that the mid-thigh sensor placement is superior. These results are supported by
243 previous research that found data from a single mid-thigh inertial sensor were a better predictor of peak
244 knee extensor moment and power absorption during a single limb task than a mid-shank sensor³⁵ and
245 that data from a single thigh accelerometer, but not a single shank accelerometer, were predictive of
246 between-limb differences in knee power absorption during running³⁶.

247 In the current study, both of the mid-segment inertial sensors were placed directly on the rigid
248 optical motion capture marker clusters used for segment tracking. However, it is the anatomical
249 coordinate system defined by markers placed on bony landmarks, rather than these rigid clusters, that
250 define the segment coordinate systems for the motion capture measures. Orientation of the inertial
251 sensors relative to the anatomical coordinate system is a key factor in accuracy of joint angles measured
252 by inertial sensors³⁷. A varying degree of curvature on the lateral aspect of the shank due to the shape of
253 the lateral head of the gastrocnemius muscle could have resulted in less consistent placement of the
254 mid-shank inertial sensor among legs, thus affecting the relationship between the inertial sensor
255 coordinate system and the anatomical coordinate system of the shank. In contrast, the lateral thigh has
256 less variation in curvature along its length, which may have ensured more consistent placement of the
257 mid-thigh inertial sensor across legs. Thus, the mid-thigh sensor placement may be preferable to the
258 mid-shank placement for assessing varus thrust in individuals with knee OA, particularly in cases where
259 limited time or experience prevent more standardized placement.

260 The regression models for EKAM also provide support for the mid-thigh versus mid-shank
261 placement as the association between segment angular velocity and EKAM was not consistent across
262 legs for the mid-shank sensor, i.e. it was significant for the left but not the right leg. While the current
263 dataset did not provide enough power for a thorough comparison between left and right legs, these
264 groups were similar in terms of KLG and HKA, thus no difference between legs was expected. Despite
265 this discrepancy between legs for the mid-shank sensor, the novel finding that mid-thigh adduction
266 velocity was significantly associated with EKAM, even after adjusting for confounders, is promising as it
267 suggests that an estimation of EKAM may be possible without the need for extensive laboratory
268 equipment or analysis of multiple inertial sensors. This finding also supports previous studies that have
269 shown associations between various quantitative measures of varus thrust and EKAM^{17,20,21}.

270 In the current study, only single-sensor inertial measures were examined with the idea that
271 large-scale screening in a clinical setting would require a quick set-up with minimal data processing.
272 However, the lack of an exact 1:1 increase in adduction velocity between these inertial sensor and
273 optical motion captures measures may be attributed to the fact that the inertial sensor segment
274 adduction velocity measures only describe movement of a single segment (thigh or shank), while knee
275 adduction velocity describes the relative movement between the thigh and shank segments. Integration
276 of multiple inertial sensors or sensor components may result in better estimation of knee adduction
277 velocity, however, this typically requires a series of functional calibration exercises, modeling
278 assumptions, and data filtering to address issues such as drift and sensor alignment³⁸. The finding in the
279 current study of a significant association between single inertial sensor measures and optical motion
280 capture measures suggests a single sensor may be sufficient as a quick screening tool for severity of
281 varus thrust in knee OA populations where more extensive data collection is not feasible.

282 We observed significant associations of mid-thigh and mid-shank sensors with both knee
283 adduction velocity and knee adduction excursion measures at both walking speeds. As noted by Chang

284 et al.⁶, knee adduction velocity captures the speed of the movement and can be used as a reliable
285 measure of varus thrust in people with knee OA. Knee adduction excursion has also been used as a
286 quantitative metric of varus thrust and has been associated with both OA severity (KLG) and EKAM in
287 individuals with medial knee OA²¹. Associations would be expected between measures of thigh/shank
288 adduction velocity from the inertial sensors and knee adduction velocity from optical motion capture
289 given the similar constructs being measured. However, significant associations of metrics from
290 thigh/shank inertial sensors and knee adduction excursion further support the use of inertial sensors to
291 assess varus thrust. Furthermore, the similarity of results across the two speeds for the analyses in the
292 current study suggests that a single sensor could be used to quantify varus thrust across different
293 walking speeds in individuals with knee OA.

294 There are a number of limitations of the current study that should be acknowledged. Given that
295 this was a proof of concept study, the sample size was small. Data error resulted in the loss of inertial
296 sensor data for some legs due to signal clipping. While this appeared to be an issue with software
297 presets, rather than the inertial sensors themselves, the loss of this data and small sample size overall
298 did not provide enough power for a definitive comparison across sensor locations or speeds. The lack of
299 standardized placement of the inertial sensors on the leg, e.g. a specified percent distance along a given
300 segment, may have increased variability in the measurement of segment adduction velocities from the
301 inertial sensors, resulting in misalignment of coordinate systems between the inertial and optical motion
302 capture systems and an inability to provide a definitive recommendation on sensor placement. The
303 attachment of sensors in a clinical setting would likely also be done without standardized placement or
304 calibration and thus these results may be a good representation of how clinical inertial sensor data
305 would correspond to optical motion capture data. It should be noted, however, that the sensors in the
306 current study were placed on top of the rigid plate containing optical motion capture markers and it is
307 unclear whether a rigid sensor alone versus the rigid sensor/plate combination would produce similar

308 data. In this study, gait events identified by the optical motion capture system were used for calculation
309 of the inertial measures. In a clinical setting where optical motion capture data are not available, gait
310 events calculated from the inertial sensors, as has been done previously³⁹, would need to be utilized.
311 Reliability of the thigh inertial sensor measurement will also need to be established before it can be
312 used in clinical settings, particularly given the limited number of studies investigating reliability of
313 inertial sensors in knee OA populations^{40,41}. It should also be noted that while the sample in the current
314 study is similar to that of larger studies of individuals with medial knee OA in terms of age, sex, and
315 BMI¹³, the results may not generalize to other sub-groups of the OA population such as individuals with
316 BMI > 40.

317

318 **5. Conclusions**

319 In this proof-of-concept study, we demonstrated a significant association between increases in thigh
320 angular velocity derived from the gyroscope signal of a single inertial sensor and increases in surrogate
321 varus thrust measures derived from an optical motion capture system. Furthermore, increased thigh
322 angular velocity from this single inertial sensor was associated with increased peak EKAM after adjusting
323 for confounders. These results highlight the potential of inertial sensors for quantifying varus thrust
324 without the need for an optical motion capture system.

325

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333

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444 treadmill walking in knee osteoarthritis patients. *Arch Phys Med Rehabil*. 2008;89(4):732-737.

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446

447

448 Table 1. Sample characteristics

Clinical & Demographic Characteristics	n = 26 participants†
Sex	n = 16 [62%] female
Age (years)	64.5 (8.4)
Body mass index (kg/m ²)	28.6 (4.7)
Height (m)	1.67 (0.12)
Gait speed (m/s):	
Self-selected	1.32 (0.22)
Fast	1.62 (0.32)
Knee Injury and Osteoarthritis Outcome Score - Pain (/100*)	59.2 (11.1)
Kellgren-Lawrence Grade (KLG):	
KLG = 2	n = 13 legs
KLG = 3	n = 26 legs
KLG = 4	n = 8 legs
Static limb alignment (degrees)**	175.9 (3.6)

Data presented as mean (standard deviation) except where noted

†n = 47 knees from these participants had KLG ≥ 2. KOOS, KLG, & alignment values are reported for these knees only.

**Lower scores represent greater pain*

***Angles < 180 degrees indicate static varus alignment*

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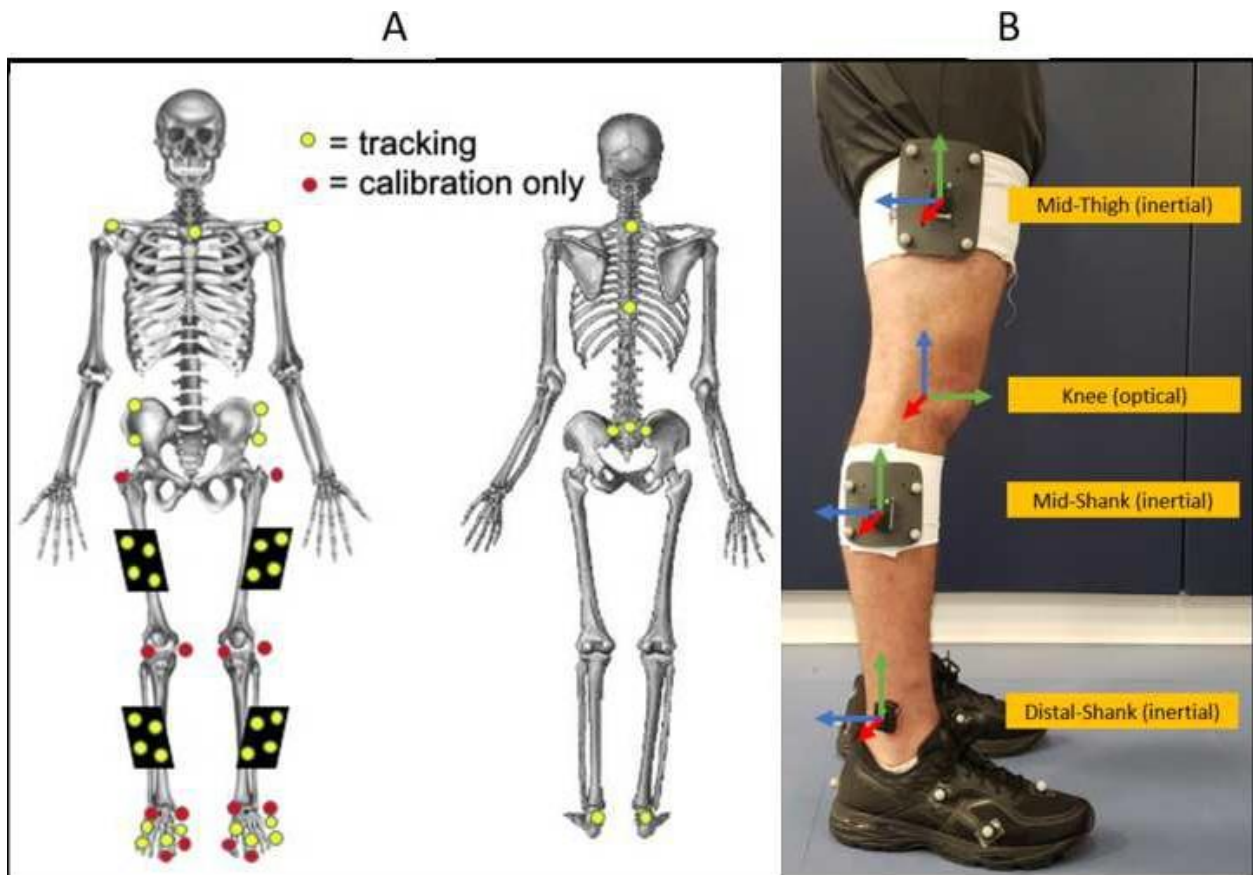
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451 Table 2: Inertial and optical motion capture outcome average values in knees with radiographic OA (KLG
 452 ≥ 2)

Inertial motion capture	
Mid-thigh adduction velocity (degree/s) (n = 39 legs):	
Self-selected speed walking	45.2 (25.2)
Fast speed walking	58.9 (28.1)
Mid-shank adduction velocity (degree/s) (n = 41 legs):	
Self-selected speed walking	-80.4 (36.9)
Fast speed walking	-97.3 (43.4)
Optical motion capture	
Knee adduction velocity (degree/s) (n = 45 legs):	
Self-selected speed walking	63.9 (24.8)
Fast speed walking	75.6 (28.6)
Knee adduction excursion (degree) (n = 45 legs):	
Self-selected speed walking	3.8 (2.0)
Fast speed walking	4.0 (2.1)
External knee adduction moment (% bodyweight-height) (n = 45 legs):	
Self-selected speed walking	3.25 (1.22)
Fast speed walking	3.71 (1.43)

Data presented as mean (standard deviation)

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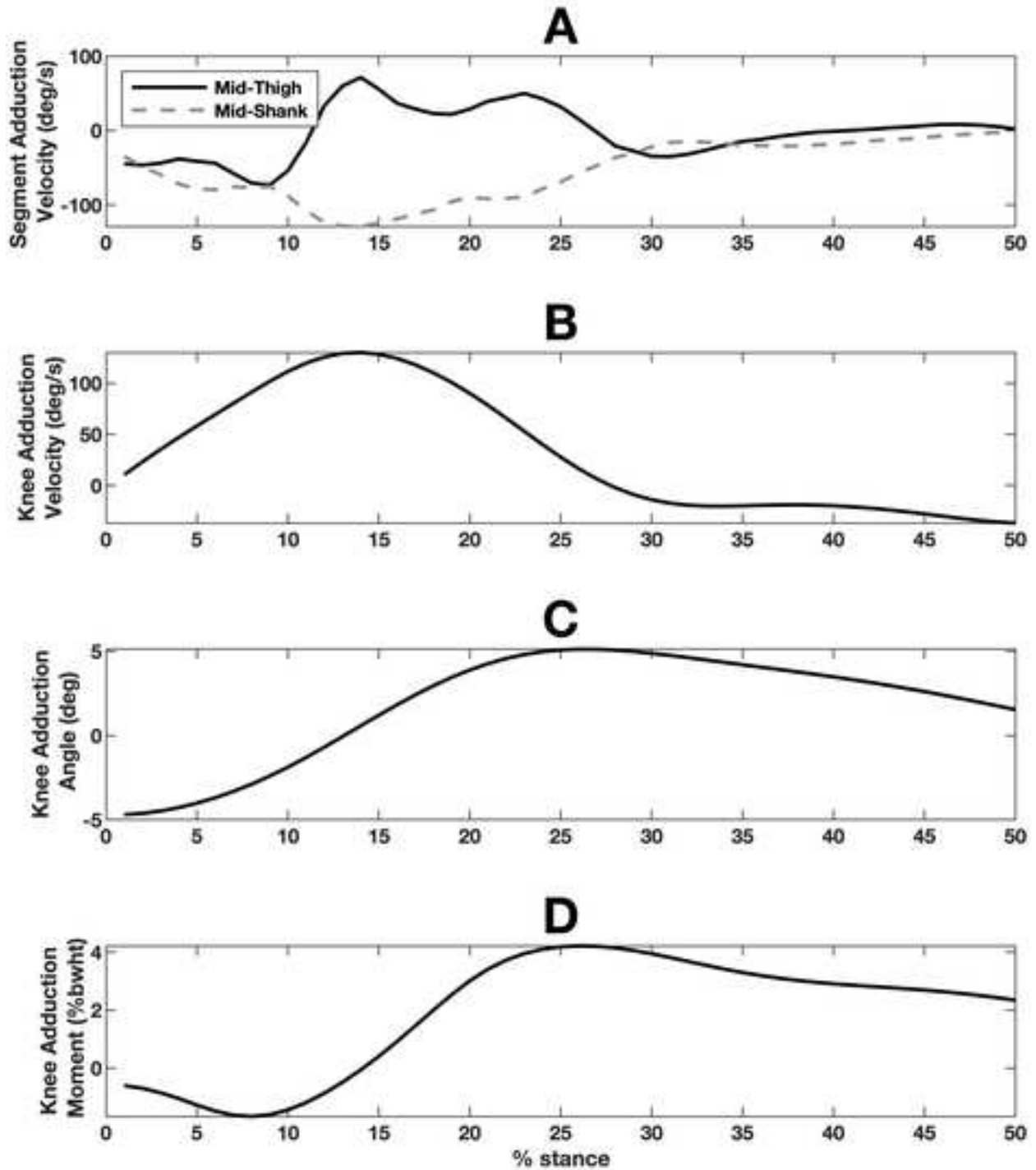
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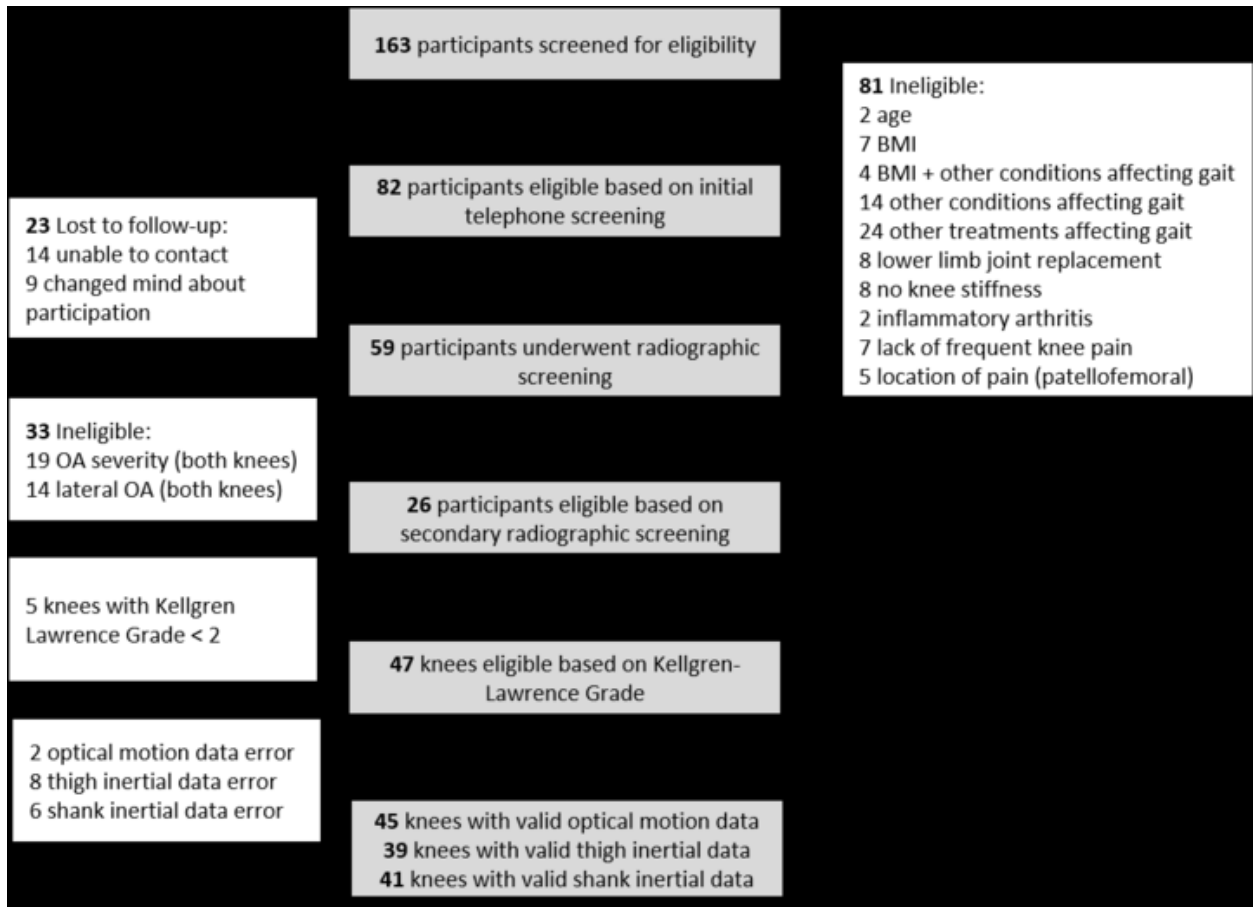
459

Figure 1. (A) Optical motion capture markers (red markers used for static trial only) and (B) inertial sensors, showing placement of the mid-thigh and mid-shank inertial sensors on the optical motion capture marker clusters for the thigh and shank segments, respectively, and distal shank inertial sensor directly on the skin on the lateral aspect of the distal tibia, along with the coordinate systems for the knee (optical system) and inertial sensors.



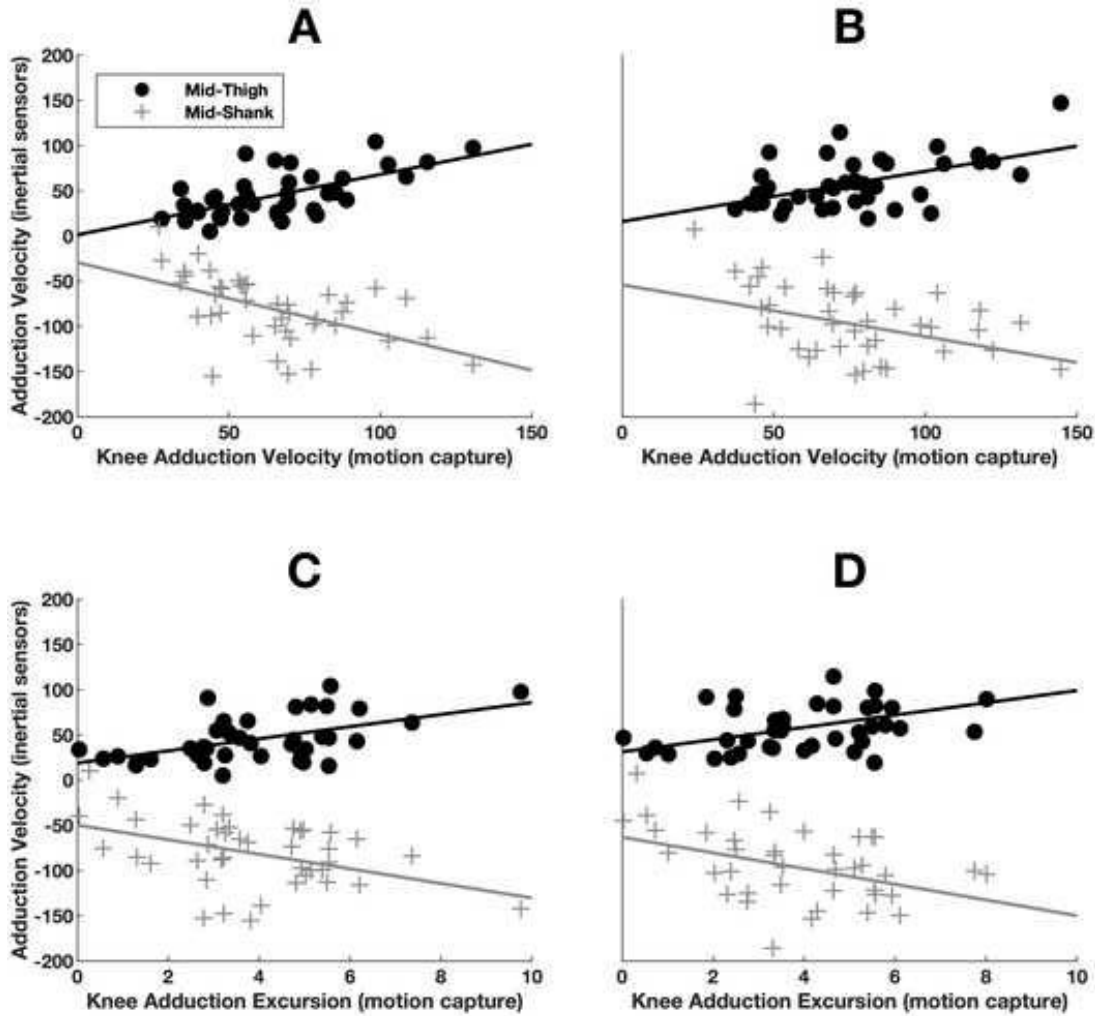
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461 Figure 2. Representative waveforms from self-selected speed walking showing (A) inertial sensor
 462 segment adduction velocity from the mid-thigh and mid-shank sensors, (B) knee adduction velocity, (C)
 463 knee adduction angle and knee adduction excursion, and (D) external knee adduction moment (EKAM).



464

465 Figure 3. Study recruitment and final study sample.



466

467 Figure 4. Relationships between segment adduction velocity (captured by the inertial sensors) and (A)
 468 knee adduction velocity during self-selected speed walking, (B) knee adduction velocity during fast
 469 speed walking, (C) knee adduction excursion during self-selected speed walking, and (D) knee adduction
 470 excursion during fast speed walking in knees with radiographic osteoarthritis.

471