

Research article

Effect of kayak ergometer elastic tension on upper limb EMG activity and 3D kinematics

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Abstract

Despite the prevalence of shoulder injury in kayakers, limited published research examining associated upper limb kinematics and recruitment patterns exists. Altered muscle recruitment patterns on-ergometer vs. on-water kayaking were recently reported, however, mechanisms underlying changes remain to be elucidated. The current study assessed the effect of ergometer recoil tension on upper limb recruitment and kinematics during the kayak stroke. Male kayakers ($n = 10$) performed 4 by 1 min on-ergometer exercise bouts at 85% VO_2max at varying elastic recoil tension; EMG, stroke force and three-dimensional 3D kinematic data were recorded. While stationary recoil forces significantly increased across investigated tensions (125% increase, $p < 0.001$), no significant differences were detected in assessed force variables during the stroke cycle. In contrast, increasing tension induced significantly higher *Anterior Deltoid* (AD) activity in the latter stages (70 to 90%) of the cycle ($p < 0.05$). No significant differences were observed across tension levels for *Triceps Brachii* or *Latissimus Dorsi*. Kinematic analysis revealed that overhead arm movements accounted for $39 \pm 16\%$ of the cycle. Elbow angle at stroke cycle onset was $144 \pm 10^\circ$; maximal elbow angle ($151 \pm 7^\circ$) occurred at $78 \pm 10\%$ into the cycle. All kinematic markers moved to a more anterior position as tension increased. No significant change in wrist marker elevation was observed, while elbow and shoulder marker elevations significantly increased across tension levels ($p < 0.05$). In conclusion, data suggested that kayakers maintained normal upper limb kinematics via additional AD recruitment despite ergometer induced recoil forces.

Key words: Kayaking, ergometry, 3D joint kinematics, electromyography, shoulder.

Introduction

Flatwater kayaking is a sport requiring a high degree of coordination and skill in order to perform at an elite level, in part due to the complex movements necessary to propel the unstable racing kayak forward (Cox, 1992). The kayak stroke cycle involves a series of contralateral movements of the upper body, with four distinct phases (an individual draw and transition phase for both right and left sides). The draw phases comprise the time intervals when the paddle is submerged in the water. In contrast, the transition phases comprise the time intervals when neither paddle is submerged (Fleming et al., 2012). This complex series of upper body movement requires a wide range of articulations and motion from the shoulder joint and upper limb.

It is therefore unsurprising that injuries sustained in

flatwater kayaking are almost entirely restricted to the upper body (Edwards, 1993; Hagemann et al., 2004). Of the upper body injuries occurring in kayaking, the greatest proportion (53%) are shoulder related (Edwards, 1993). In order to better understand the mechanisms underlying this high injury prevalence, several papers have assessed both upper limb muscle recruitment (Trevithick et al., 2007), three-dimensional (3D) body segment (Begon et al., 2008) and scapulohumeral kinematics (Wassinger et al., 2011) during the kayak stroke cycle. While this literature has added greatly to our understanding of the biomechanical demands of kayaking, all data were recorded during ergometer kayaking, which may not accurately represent the on-water scenario (Mitchell and Swaine, 1998; Van Someren et al., 2000).

A recently published paper reported significant differences in upper limb muscle activity comparing on-ergometer and on-water kayaking (Fleming et al., 2012). These authors hypothesised that the observed differences were due to additional recoil forces acting upon the ergometer paddle shaft. During ergometer kayaking, appropriate tension must be maintained between the paddle and flywheel via the connecting chords; this is accomplished via the ergometer's adjustable elastic tension. However, it has been suggested that this elastic tension significantly alters the normal muscle recruitment patterns recorded during on-water kayaking (Fleming et al., 2012). The current study's primary aim was to test this hypothesis, by analysing the effect of the ergometer's elastic tension on both muscle activity and 3D upper body joint kinematics during the kayak stroke cycle. By increasing the elastic tension applied to the paddle shaft and comparing joint kinematics and associated EMG activity during matched bouts of fixed intensity kayaking, any independent effects of this external recoil force could be elucidated.

In order to maintain optimal stroke biomechanics, the kayaker may resist external forces via altered recruitment of shoulder and arm musculature; conversely, the external forces applied on-ergometer may directly alter normal upper limb kinematics, thus affecting associated muscle recruitment patterns. It remains to be seen whether differences in EMG observed comparing on-ergometer and on-water kayaking are the result of efforts (conscious or otherwise) to maintain optimal movement patterns, or if the external forces are acting upon joints to create both altered movement and EMG patterns. A secondary aim was therefore to quantify shoulder, scapular, elbow and wrist kinematics during the on-ergometer stroke cycle. We hypothesised that increasing the ergometer's elastic

recoil tension would result in increased EMG activity, without significant changes in joint kinematics.

Methods

Participants

Ten ($n = 10$) male international flatwater kayakers volunteered (mean \pm SD; age 21 ± 3 yr, height 1.80 ± 0.06 m, body mass 74.6 ± 5.8 kg). All had previously performed both maximal graded incremental tests and task specificity trials both on-ergometer and on-water (Fleming et al., 2012). All participants were fully informed of procedures involved in the current study and provided written informed consent. Ethical approval for this study was granted from the University Health Sciences ethics committee.

Experimental design

This study design consisted of two visits. Initially, a graded incremental test to volitional exhaustion was performed on a Dansprint kayak ergometer (Dansprint, Hvidovre, Denmark) to assess VO_2 , lactate and heart rate response profiles (Fleming et al., 2012). Data acquired during incremental testing were subsequently used to set each kayaker's exercise intensity ($85\%VO_{2peak}$) for their 3D kinematic exercise trial. Elastic recoil tension, applied by the ergometer's loading mechanism, was adjusted during the kinematic trial in order to assess the independent effect of applied recoil forces on muscle activity and joint kinematics of the shoulder joint and upper limb.

Kinematic trial

A 10-min warm-up at power output equivalent to 50% of each individual's VO_{2peak} was performed prior to commencing 3D kinematic trials. The trial itself consisted of a series of 4 by 1 min matched exercise bouts; these exercise bouts were preceded by a 3 min rest period in order to eliminate any risk of fatigue affecting subsequent bouts. Elastic tension applied to the ergometer's paddle shaft via the connecting pulleys was varied between each exercise bout. Tension was increased or decreased in a stepwise fashion via shortening or lengthening of the ergometer's elastic chord by fixed lengths of 10% relative to the overall chord length for each respective exercise bout. The elastic tensions ranged from 0% shortened (T1) up to 30% shortened (T4). Trials were randomised into ascending or descending order, to eliminate possible effect of tension order on muscle activity. Throughout the entire trial, kayakers maintained a fixed power output (W) and stroke rate ($strokes \cdot min^{-1}$) equivalent to 85% of their VO_{2peak} . This was achieved via the ergometer's LCD screen which continually provided visual feedback of both power output and stroke rate. Since the exercise duration was significantly shorter (1 vs. 3 min) than that used previously (Fleming et al., 2012) power output instead of heart rate was considered a more appropriate means of quantifying exercise intensity. By fixing both power output and stroke rate, it was envisaged that the effect of ergometer elastic tension could be assessed independently of all other variables.

EMG and stroke force data

EMG data were recorded from *Triceps Brachii* (long head) (TB), *Anterior Deltoid* (AD) and *Latissimus Dorsi* (LD) on the right side of the body (ME6000 Mega, Koupio, Finland). Synchronisation of EMG and video data using an audio-sync trigger (Mega, Koupio, Finland) facilitated identification of onset of each stroke cycle on the EMG recording. EMG data from 10 consecutive stroke cycles in the latter stages of each tension trial were amplitude processed via root mean squaring and normalised relative to pre-trial isometric MVC, respectively (Fleming et al., 2012). Subsequent temporal normalisation and averaging via cubic spline fitting produced an average rmsEMG ensemble for each muscle during kinematic trials. Stroke force data were recorded using strain gauge arrays integrated into a carbon paddle shaft as previously described (Fleming et al., 2012).

3-D kinematic data

Movement patterns of discrete anatomical reference points were recorded using a CODA dual CX1 motion analysis system (Charnwood Dynamics, Rothley, UK). The CODA motion analysis system used active infra-red LED markers to measure positions within a 2 by 2 by 3 m³ volume. The translational precision of the instrument has been reported to be within 0.5 mm in each direction, while rotational accuracy is within 1°, determined using factory calibration experiments (Mottram et al., 2009). Two separate CX1 measurement units were placed equidistant (approximately 5m) and orthogonally to the left and right sides of the sagittal plane. Prior to motion capture, the CODA system was pre-calibrated by placing fixed reference points on the ground within the measurement volume. Marker positions were captured at 100 Hz and data transferred to PC for subsequent analysis using CODA software (CODAmotion V2.0, Charnwood Dynamics, Rothley, UK). All 3D kinematic data were presented in the horizontal (X), lateral (Y) and vertical (Z) axes as displacement in mm from the pre-calibrated reference points. Since the ergometer was positioned behind the calibrated reference points, all marker positions were recorded as negative displacements in the X-axis, positive displacements in the Z-axis and both positive and negative displacements in the Y-axis, depending on their relative position during the kayak stroke.

Previous studies have reported that skin mounted motion sensors are suitable to measure scapula rotation and translation (Zhou et al., 2009; Mottram et al., 2009). While the accuracy of all skin mounted marker-tracking systems is inherently limited (Mottram et al., 2009), accuracy was deemed satisfactory for the purposes of the current study. Markers were attached to the head of the *Ulna* (wrist marker) and the lateral epicondyle of the *Humerus* (elbow marker) in order to assess movement patterns of the right arm. Additional markers were attached to the lateral tip of the *Acromion* (shoulder marker), *Inferior Scapula* and medial border of the *Spinal Scapula* (scapular markers). Marker locations were identified via anatomical reference to bony prominences and all markers were secured to the skin with double-sided adhesive tape (Sellotape, Cheshire, UK).

Data reduction and statistical analysis

All EMG, force and 3D motion data were transferred to Matlab (Mathworks, Massachusetts, USA) for subsequent data reduction. For statistical analysis, mean rmsEMG data were averaged for each 10% interval of the stroke cycle. Stroke force data were also averaged over the same 10 consecutive stroke cycles and temporally normalised to attain a group mean stroke force ensemble at each 2% of stroke cycle duration. Data were subsequently analysed to attain measures of peak force (N), absolute time to peak force (s), normalised time to peak force (%), rate of peak force development (RFD_{peak} in $N \cdot s^{-1}$) and rate of 50% peak force development (RFD_{50} in $N \cdot s^{-1}$) as outlined by Benson et al. (2011). Integration of the stroke force profile in the first 30% of the stroke cycle quantified the draw impulse (N.s). 3D kinematic movement patterns for each anatomical marker were temporally normalised and averaged over the same 10 consecutive stroke cycles. In addition, 3D kinematic data were then averaged for each 10% interval of the stroke cycle to facilitate statistical analyses at comparative phases relative to the EMG data. Elbow joint angle was quantified throughout the cycle. Overhead arm movement was quantified as any movement in which elbow marker elevation exceeded that of the acromion. For the purposes of describing shoulder and arm kinematics during the stroke cycle, data from the lowest tension level (T1) was used, as this represented the closest scenario to on-water kayaking within the mechanical limitations of the ergometer.

Data are presented as group mean \pm SD unless otherwise stated. Normality of all data sets was assessed using Kolmogorov-Smirnov tests. Statistical analyses were performed using 1-way repeated measures ANOVA; *post-hoc* Tukey tests quantified detected differences.

Statistical analyses were performed using Sigma Stat (Systat Software, Chicago, USA) with $p < 0.05$ infer-

ring statistical significance.

Results

EMG data

Group mean (SD) data for iEMG are presented in Table 1, no significant differences were observed for iEMG data in any of the muscles investigated. However, when data were normalised to MVC and averaged over 10% intervals, significant differences were observed in AD activity at discrete phases of the stroke cycle. Mean AD activity was significantly lower at T4 vs. T1 during the 40% interval ($p < 0.05$, see Figure 1), indicating that an increase in recoil force reduced activity during this phase of the stroke cycle. The opposite effect was observed during later phases, where mean AD activity was significantly greater as tension increased during the 70, 80 and 90% intervals of the stroke cycle. Mean normalised AD activity was significantly greater comparing T4 to T1 and T2 during the 70% (18.1 ± 12.0 vs. $10.6 \pm 5.3\%$, $p < 0.05$; vs. $11.7 \pm 6.1\%$, $p < 0.05$), 80% (37.6 ± 16.2 vs. $22.7 \pm 11.0\%$, $p < 0.001$; vs. $29.9 \pm 13.2\%$, $p < 0.05$) and 90% intervals (26.3 ± 10.1 vs. $15.6 \pm 8.9\%$, $p < 0.001$; vs. $19.8 \pm 8.6\%$, $p < 0.05$), see Figure 1. In addition, mean normalised AD activity at T3 was significantly greater ($p < 0.01$) than T1 during both the 80 and 90% intervals (32.3 ± 11.7 vs. 22.7 ± 11.0 and 24.1 ± 10.5 vs. $15.6 \pm 8.9\%$, respectively). No significant differences in muscle activity for TB and LD were observed during any discrete 10% interval of the stroke cycle.

Stroke force data

Group mean (SD) stroke force data are presented in Table 1. No significant differences were observed for peak force, time to peak force, normalised time to peak force, RFD_{peak} , RFD_{50} or stroke impulse. Significant differences

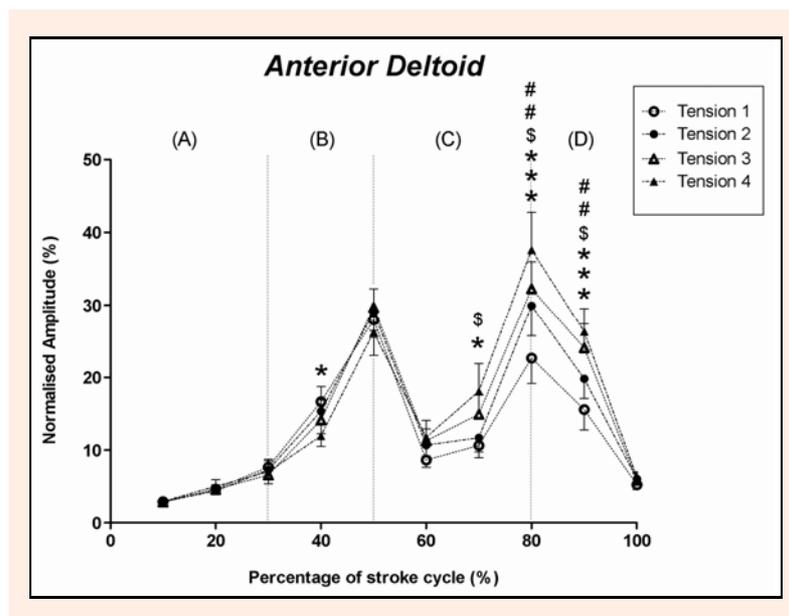


Figure 1. Group mean \pm SEM ($n=10$) normalised EMG amplitude for AD at each 10% interval of the kayak stroke cycle across all tension levels. Asterisk infer significant difference between T1 and T4 ($* p < 0.05$, $*** p < 0.001$). Hash infer significant difference between T1 and T3 ($## p < 0.01$). Dollar infer significant difference between T2 and T4 ($\$ p < 0.05$). Approximate phases of the kayak stroke cycle are represented by A (draw), B (transition), C (opposite draw) and D (opposite transition).

Table 1. Group mean (SD) iEMG and stroke force data.

	Tension 1	Tension 2	Tension 3	Tension 4
EMG activity (n=10)				
iEMG of <i>TB</i> ($\mu\text{V}\cdot\text{s}$)	161 (57)	157 (48)	164 (41)	170 (45)
iEMG of <i>LD</i> ($\mu\text{V}\cdot\text{s}$)	163 (48)	163 (49)	163 (49)	164 (49)
iEMG of <i>AD</i> ($\mu\text{V}\cdot\text{s}$)	402 (139)	437 (137)	458 (140)	466 (186)
Stroke force (n=9)				
Peak force (N)	270 (40)	282 (50)	279 (42)	286 (48)
Time to peak (s)	0.15 (0.02)	0.15 (0.03)	0.15 (0.03)	0.15 (0.03)
Time to peak (%)	9.71 (1.36)	9.64 (1.81)	9.67 (1.45)	9.59 (1.46)
RFD _{peak} ($\text{N}\cdot\text{s}^{-1}$)	1868 (398)	2025 (590)	2000 (526)	2065 (602)
RFD ₅₀ ($\text{N}\cdot\text{s}^{-1}$)	1611 (429)	1765 (421)	1716 (382)	1759 (441)
Impulse ($\text{N}\cdot\text{s}$)	66 (9)	68 (7)	72 (7)	70 (6)
Stationary recoil force (N)	20 (4) *†	29 (5) *†	37 (7) *	45 (8) †

* $p < 0.001$ compared to T4, † $p < 0.001$ compared to T3.

in the stationary forces applied to the paddle shaft were observed across all tension levels, see Table 1. Forces were significantly higher at T4 compared to T1, T2 and T3 ($p < 0.001$) and at T3 compared to T1 and T2 ($p < 0.001$). This outcome was expected, since increasing elastic tension via the progressive shortening of the elastic chord, should theoretically increase the forces applied to the paddle shaft via the ergometer pulleys.

3D kinematic data

Range of movement (ROM) plots for each marker in both the sagittal and coronal planes are presented in Figures 2 and 3. In addition, elbow joint angle throughout the stroke cycle is presented in Figure 4. Statistical analysis of the 3D marker data averaged over 10% intervals revealed significant differences across ergometer tension levels at discrete phases of the stroke cycle and within specific axes of orientation. Significant differences in marker

position relative to the horizontal axis (X-axis) were observed for each of the markers at discrete intervals of the stroke cycle. In all cases, marker position at T4 was significantly greater than during T1 or T2, inferring a more anterior position relative to the kayaker’s orientation (Figure 2). Significant differences in marker position relative to the vertical axis (Z-axis) were also observed in the elbow (*Lateral Epicondyle of Humerus*), shoulder (*Acromion*) and scapular (*Inferior Scapula*) markers at discrete intervals of the stroke cycle (Figures 2a, 2c and 2d, respectively). In all cases, these markers were significantly higher during T4 when compared to T1 or T2.

No significant differences in elevation were observed for the wrist marker (Figures 2b and 3b). No significant differences in kinematic data were observed in the lateral axis (Y-axis) for any of the markers investigated. With respect to the horizontal plane (X-

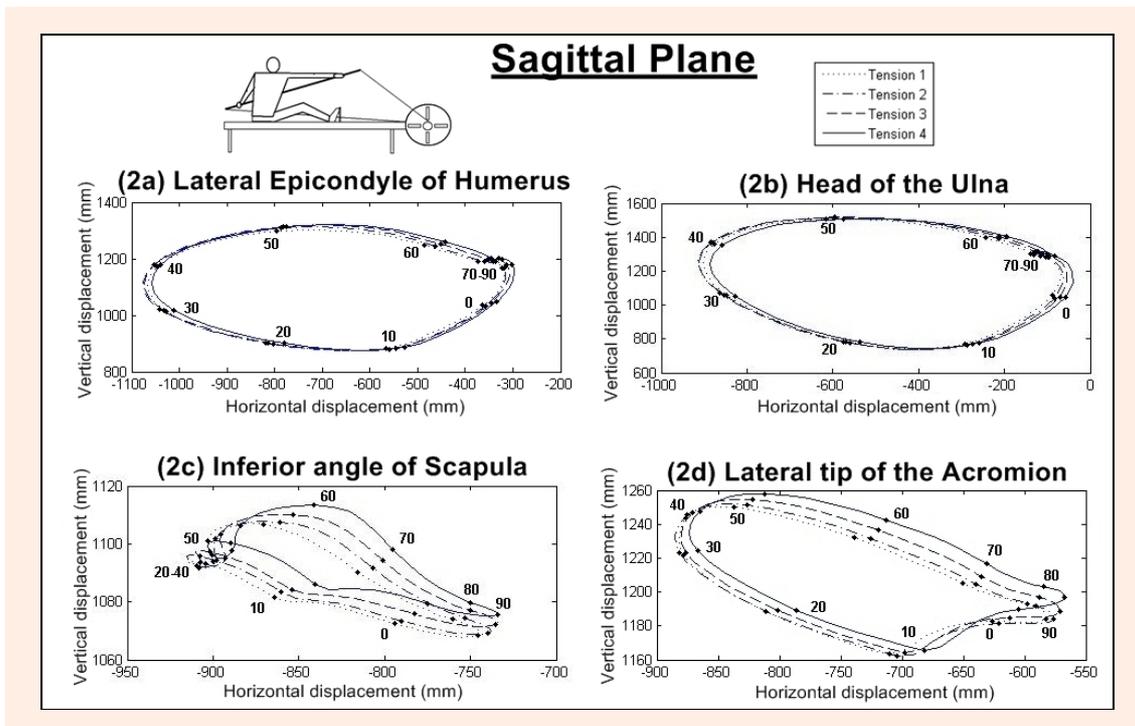


Figure 2. Kinematic data for elbow (2a), wrist (2b), scapular (2c) and shoulder (2d) markers in the sagittal plane. Data presented as group mean marker displacement (mm) at each 2% interval of stroke cycle, separate lines represent kinematic data for tensions investigated. The 10% timepoints in the stroke cycle are marked (black dot) and numbered on each trace. In addition, the spatial orientation of the kayaker is graphically depicted.

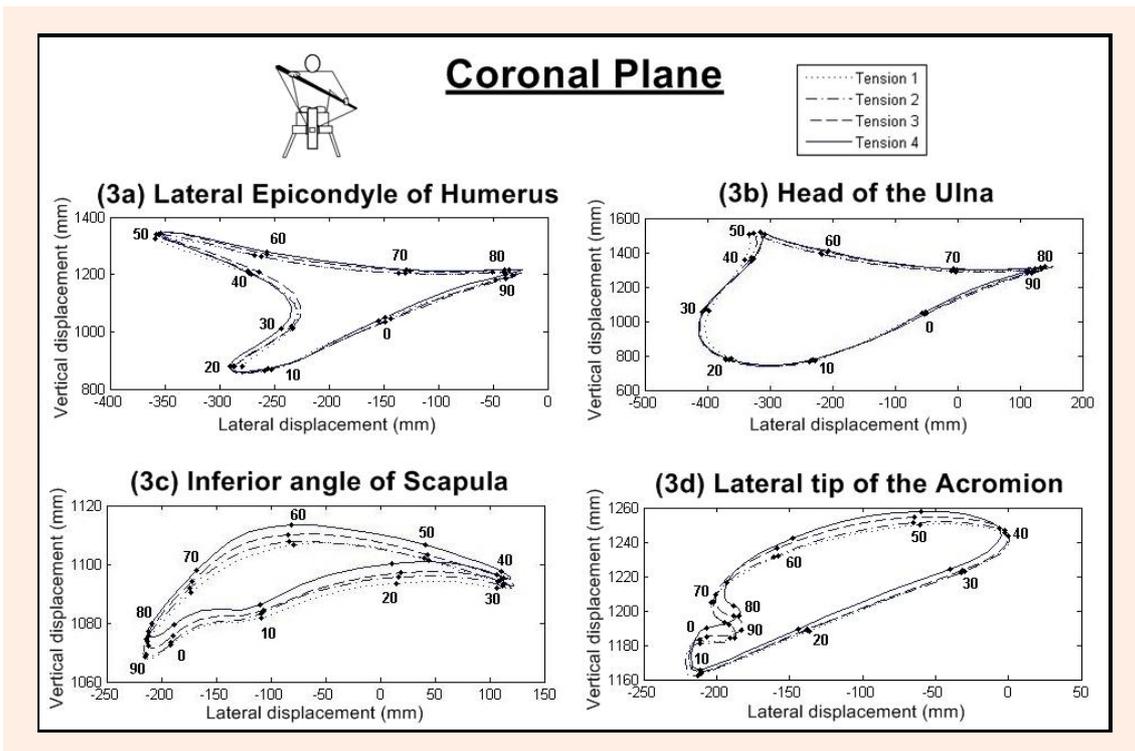


Figure 3. Kinematic data for elbow (3a), wrist (3b), scapular (3c) and shoulder (3d) markers in the coronal plane. Data presented as group mean marker displacement (mm) at each 2% interval of stroke cycle, separate lines represent kinematic data for tensions investigated. The 10% timepoints in the stroke cycle are marked (black dot) and numbered on each trace. In addition, the spatial orientation of the kayaker is graphically depicted.

axis); the wrist marker moved to a significantly more anterior position during the 10 to 40% and 60 to 100% phases of the cycle as tension increased (T4 vs. T1, $p < 0.05$). This effect was also observed in elbow marker position during the 10 to 20% (T4 vs. T1, $p < 0.01$; T4 vs. T2, $p < 0.05$) and 70 to 100% phases of the cycle as tension increased (T4 vs. T1, $p < 0.05$). As tension increased, shoulder and scapular marker positions were all in a significantly more anterior position during the 10 to

30% (T4 vs. T1, $p < 0.01$; T4 vs. T2, $p < 0.05$), 50 to 70% (T4 vs. T1, $p < 0.05$) and 90 to 100% phases of the cycle (T4 vs. T1, $p < 0.05$).

With respect to the vertical plane (Z-axis), no significant differences in kinematic data were observed for the wrist marker. Elbow marker positions were significantly higher during the 60, 70 and 100% phases of the cycle and shoulder marker position was significantly higher during the 40 to 100% phase of the cycle as

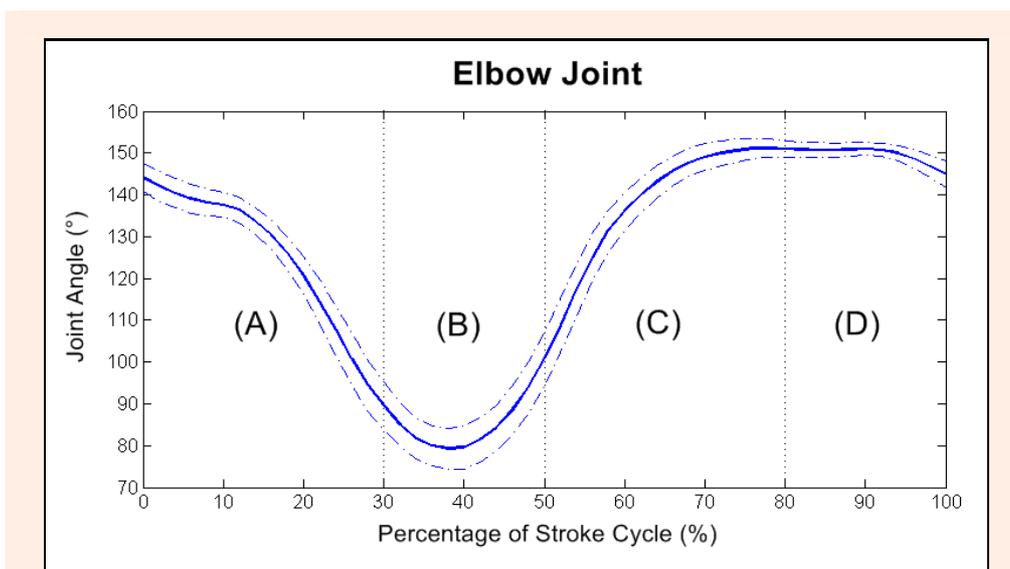


Figure 4. Elbow joint angles during the on-ergometer kayak stroke cycle. Data presented as group mean \pm SEM ($^{\circ}$) at each 2% interval of the stroke cycle during kayaking at tension level 1 (T1). The approximate phases of the stroke cycle are represented as A (draw), B (transition), C (opposite draw) and D (opposite transition).

tension increased ($p < 0.05$, Figures 2a and 2c). Scapular marker positions were significantly higher at all phases of the cycle as elastic recoil tension increased (Figure 2d).

Elbow joint angle at stroke cycle onset was $144 \pm 10^\circ$, while maximum elbow joint angle ($151 \pm 7^\circ$) occurred towards the end of the opposite draw phase of the stroke cycle (at 78%), see Figure 4. This elbow angle was maintained close to maximal for much of the opposite transition phase (80 to 100% of stroke cycle), prior to the subsequent stroke cycle onset, see Figure 4. Overhead arm movement (defined as elbow marker elevation exceeding shoulder marker elevation) accounted for $39 \pm 16\%$ of the stroke cycle. Initiation and cessation of overhead arm movement occurred at 44 ± 16 and $83 \pm 9\%$ of the stroke cycle, respectively.

Discussion

A significant difference in AD recruitment pattern, comparing on-ergometer and on-water kayaking has recently been reported (Fleming et al., 2012). This difference manifest itself as a significant second phase of AD activity during on-ergometer kayaking which was not clearly evident during the on-water scenario. This additional phase of AD activity occurred during 70, 80 and 90% intervals of the stroke cycle. It was hypothesised that additional AD activity observed during this phase of the cycle resulted from efforts to resist the downward forces acting upon the arm and shoulder. The primary aim of the current study was to assess the extent to which this external force applied by the ergometer's loading mechanism affected upper limb EMG and 3D kinematics. Additionally, upper limb joint kinematics were quantified in an effort to more clearly describe the specific patterns of movement which these joints undergo during the on-ergometer stroke cycle.

The results of the current study confirm that recoil forces associated with elastic tension were responsible for this significant second phase of AD activity observed during on-ergometer kayaking (Fleming et al., 2012). Mean AD activity was significantly higher at the 70, 80 and 90% intervals of the stroke cycle as the elastic tension was increased (Figure 1). But were these increases in AD activity resultant from the kayaker's effort to maintain optimal joint kinematics or as a result of markedly altered kinematics? The shoulder is in its most forward flexed position during the 70, 80 and 90% intervals of the stroke cycle (see Figure 2). In addition, the ergometer recoil forces acting upon the paddle shaft during this phase of the stroke cycle are in a downward trajectory. Therefore the most likely changes to joint kinematics during this phase of the cycle would be reduced forward flexion and a lowering of the arm from its normal height. Kinematic data confirmed that as elastic tension was increased, no significant changes in wrist marker elevation (vertical displacement) occurred during this phase of the stroke cycle (see Figure 2a). Furthermore, an increase in elbow marker elevation was observed during the 60 and 70% phases of the stroke cycle. Shoulder marker elevation during the latter phases of the stroke cycle was also significantly increased as tension increased (see Figure

2d). Collectively these kinematic and EMG data suggest that kayakers were actively resisting the downward recoil force via increased AD activity, in order to maintain optimal wrist position during the latter stages of the stroke cycle. At higher levels of external recoil force (such as those experienced at T3 and T4) the increases in AD activity actually raised the height of shoulder and elbow markers while the corresponding height of the wrist marker remained unchanged.

Elastic tension had the opposite effect on AD activity during the 40% interval of the stroke cycle, which is concurrent with the transition phase of the kayak stroke cycle. During this phase (30 to 50%), the shoulder is undergoing abduction as the kayaker raises the elbow and wrist (Figures 3a and 3b, respectively) in preparation for the opposite draw phase. Mean AD activity at 40% of the stroke cycle was significantly lower during T4 when compared to T1 (Figure 1). At this phase of the stroke cycle, external recoil forces are acting to pull the paddle shaft forward as the kayaker abducts their shoulder through the transition phase. Kinematic data from all markers showed significant anterior shifts in horizontal displacement throughout the stroke. As such, it seems likely that any resistance to this forward pull during the transition phase would be achieved via increased horizontal abduction and/or horizontal extension of the shoulder. This would most likely be achieved by an increase in *Posterior Deltoid*, *Infraspinatus* and *Teres Minor* activity (Hintermeister et al., 1998) during the transition phase, which may explain the reduced reliance on AD activity as elastic tension increased during the 40% interval of the cycle. No significant changes were detected in either TB or LD activity as elastic tension was increased. These findings would suggest that the recoil forces applied by ergometer's loading mechanism do not contribute to the altered EMG response observed when compared to on-water kayaking (Fleming et al., 2012). It is worth noting however, that even at T1, a significant level of external force (20 ± 4 N) was still being applied to the paddle shaft. It was not possible to lower this force any further due to the ergometer design. As such, the simple presence of an additional external force cannot be ruled out as a possible explanation for the differences in TB and LD activity previously reported when comparing on-ergometer on-water kayaking.

The external forces applied by the ergometer's elastic tension will always be directed to the anchor point (ergometer flywheel), however during the dynamic kayaking movement, the vector which the force acts is constantly changing. At certain phases of the stroke cycle, the forces are acting in a downward direction. This was highlighted in the current study during the 60 to 90% phase of the cycle, where these greater downward external forces resulted in significant increases in AD activity. Nonetheless, due to the position of the ergometer flywheel relative to the paddle shaft, the overall effect of the elastic tension is a forward recoil force. This undoubtedly explains the significant changes in kinematic data relative to the horizontal axis observed in all markers in the current study. As tension increased, all joint positions were pulled into a more anterior position

relative to the horizontal axis. The exact mechanism as to how this was achieved remains to be elucidated. It is possible that the changes observed were due to increased protraction throughout the stroke cycle. Certainly, data from the scapular and shoulder markers would infer possible protraction (Ludewig and Reynolds, 2009). However, it is also possible that a progressive change in seating position via increased hip and lumbar flexion could have resulted in a similar kinematic outcome. Since no markers were applied to either the hip, trunk or clavice, it remains to be evaluated if the consistent increases in marker position relative to the horizontal axis were a result of increased clavicular protraction or hip and/or lumbar flexion. Additional 3D kinematic analysis is clearly warranted in order to clarify this issue.

Analysis of the upper limb kinematics during the kayak stroke cycle revealed several findings of potential importance for training, biomechanical performance and injury prevention. Firstly the rate of elbow flexion during the draw phase of the stroke cycle appears non-linear. During the early phase of the draw (0-15%), the rate of elbow flexion is slow (Figure 4). Substantial TB activity has previously been reported in this phase (0-15%) during both on-water and on-ergometer kayaking (Fleming et al., 2012). During the second half of the draw phase, when TB is inactive, the rate of elbow flexion is higher (Figure 4). It therefore appears that elbow flexion is actively resisted via eccentric TB recruitment, during the first half of the draw phase. This finding is in agreement with previous literature reporting that inhibition of the elbow flexors by skilled kayakers enhanced their potential force generating capacity during the draw phase of a simulated kayak stroke (Tokuhara et al., 1987). The elbow joint angle at the onset of the stroke cycle was lower than the maximal angle recorded ($144 \pm 10^\circ$ vs. $151 \pm 7^\circ$, respectively). This finding highlighted that elbow flexion was already initiated prior to the stroke cycle onset. Cox et al. (1992) previously stated that elbow extension and trunk rotation should be maximised at stroke cycle onset, in order to enhance forward reach and stroke length. The reductions from maximal elbow angle observed at stroke cycle onset could therefore be interpreted as a manifestation of flawed stroke technique. The two top performers (based on personal best times) in the current study both exhibited little or no reductions from maximal elbow angle at stroke cycle onset, adding further support to this hypothesis.

Kinematic data also revealed that while the shoulder joint undergoes internal rotation during the early draw phase (0-20%, see Figures 3a and 3b), this articulation shifts to external rotation as the draw phase concluded. This shift in articulation is clearly observed as a rapid internal movement of the elbow marker despite continuing external movement of the wrist marker in the coronal plane (see Figures 3a and 3b, 20-30%).

The final finding of note was that a high proportion of the stroke cycle ($39 \pm 16\%$) was performed with the elbow marker elevation exceeding shoulder elevation, inferring that the shoulder spends a significant period of time in a state of abduction or flexion (during overhead movement). This period coincides with the contralateral

draw phase (opposite draw), in agreement with recent data from Wassinger et al. (2011). The current study however, reports peak humeral elevation occurring at 50% stroke cycle duration (see Figure 3a and 3b), in contrast to previous data reporting minimal humeral elevation at the same timepoint (Wassinger et al., 2011). This discrepancy is most likely due to differences in forward stroke technique between flatwater kayakers (used in the present study) and a cohort of whitewater kayakers previously used (Wassinger et al., 2011). Nonetheless, the significant period of overhead movement which occurs during the kayak stroke may in part explain the high prevalence of shoulder injuries which have previously been reported in kayaking populations (Edwards, 1993; Hagemann et al., 2004). Sports involving repetitive overhead arm movements are highly prone to overuse shoulder injuries (Conte et al., 2001; Cools et al., 2005; Wilk et al., 2009). Considering this literature, the current findings would suggest that more attention may be required to improving scapular and glenohumeral functional strength and control in order to reduce the risk of shoulder injury in kayakers.

Finally, it is worth noting that during training, the kayak ergometer's elastic tension is set to an individual's preference. Feedback from the current study suggests that the normal range of elastic tension is between T1 and T2. The higher tension levels (T3 and T4) would never be utilised during normal training or testing scenarios, however, for the purposes of identifying the biomechanical effect of elastic recoil force, it was necessary to include these in the current protocol. The findings of this study and previously published data (Fleming et al., 2012) show that recruitment patterns in AD are being significantly altered even at low ergometer recoil tensions (T1 to T2). This suggests that a strategy of applying minimal ergometer recoil tension during training is the best approach for a more accurate replication of the true on-water scenario.

Conclusion

From the results of the current study, the kayak ergometer's built-in loading mechanism appears to be responsible for the significant second phase of AD activity observed during the latter stages of the on-ergometer stroke cycle. When the elastic tension was increased, mean AD activity during this phase progressively increased (Figure 1). In addition, it seems likely that these increases in AD activity are as a result of the kayaker's efforts to maintain optimal joint kinematics during this phase of the stroke cycle. The fact that wrist marker elevation remained unchanged despite increasing downward forces suggests that the kayaker strives to maintain optimal hand position during the latter stages of the cycle and will alter shoulder muscle activity in response to an external force, in order to achieve this goal. The lack of any significant changes in TB and LD activity as tension was increased suggests that the recoil forces associated with the ergometer's elastic tension, do not play as significant a role in altering recruitment patterns in these muscles during on-ergometer kayaking, as was initially hypothesised. Finally, analysis of the overall

kinematic data revealed that overhead movements account for a high proportion ($39 \pm 16\%$) of the stroke cycle, and elbow flexion during the early draw phase is actively resisted via TB activity. Both these findings may have implications for strength training and technical coaching of flatwater kayakers.

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Key points

- Kayak ergometer elastic tension significantly alters *Anterior Deltoid* recruitment patterns.
- Kayakers maintain optimal arm kinematics despite changing external forces via altered shoulder muscle recruitment.
- Overhead arm movements account for a high proportion of the kayak stroke cycle.

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