Evaluating MRI and Optical Marker Tracking for Deformation Measurement in the Biofidelic ANGUS Phantom under Mild Head Impacts

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INTRODUCTION

- Traumatic brain injury (TBI) affects both military and civilian populations; however, no validated platforms currently bridge phantom-based and *in vivo* measurements to establish objective injury thresholds.
- Injurious loading cannot be applied directly to the living human brain; therefore, biofidelic brain phantoms are appealing for studying brain biomechanics.
- The Anthropomorphic Neurologic Gyrencephalic Unified Standard (ANGUS) phantom was developed to replicate key mechanical properties and representative gyral and sulcal anatomy to provide a more realistic biomechanical response under impact [1,2].
- It has been characterized using optical tracking [1] and tagged MRI (tMRI) [2] under independent injurious and non-injurious loading conditions, respectively.
- To correlate optical tracking and tMRI measurements, the ANGUS phantom was evaluated under identical loading conditions using both methods.

METHODS

- The ANGUS phantom utilizes a 3D-printed polylactic acid skull enclosing a polyacrylamide–linear acrylamide hydrogel with anatomical features, including cortical gyri, ventricles, and the falx (Fig. 1a, 1b).
- A mid-plane layer of micron-sized particles was embedded as optical tracers to track internal motion.
- Loading conditions simulating neck rotation and extension were captured using a high-speed camera (1,000 fps) and quantified via particle image velocimetry [1] (Fig. 1c, 1d).
- tMRI was performed on a central slice under the same loading conditions [2] (Fig. 2, 3).
- Both methods produced 2D displacement fields, from which Lagrangian strain tensors were estimated (Fig. 4).
- Angular acceleration and maximum principal strain (MPS) were compared across the two methods and with previously reported *in vivo* data (Fig. 5, 6).

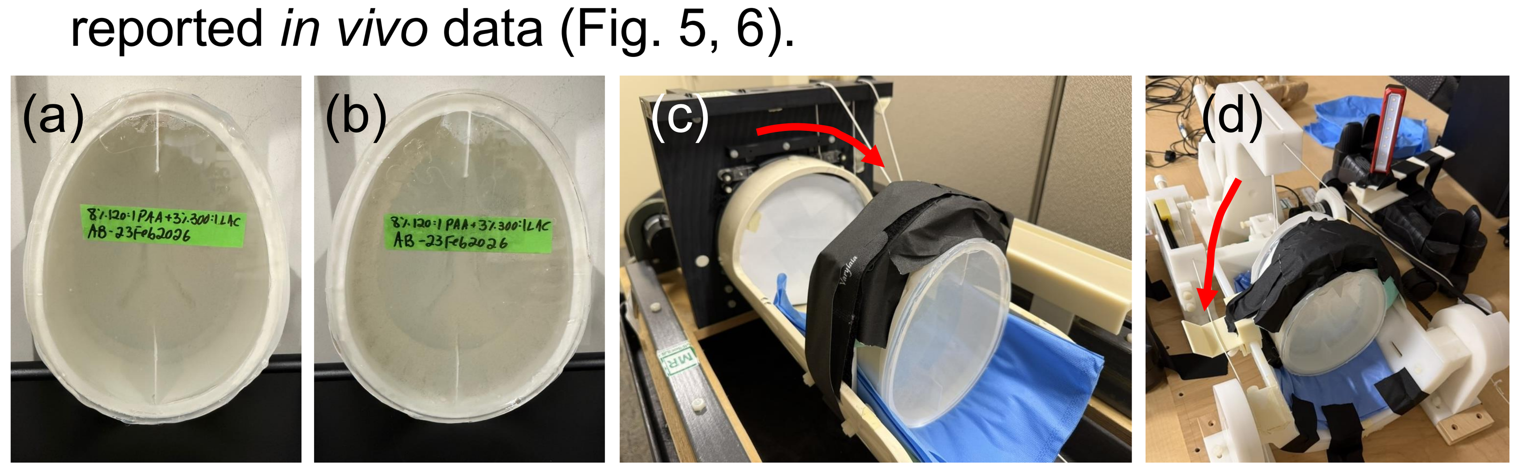

Figure 1. (a, b) Two identical ANGUS phantoms. (c) ANGUS phantom mounted in the neck rotation device (about 32º). (d) ANGUS phantom mounted in the neck extension device (about 5º). The red arrow indicates the direction of motion.

RESULTS

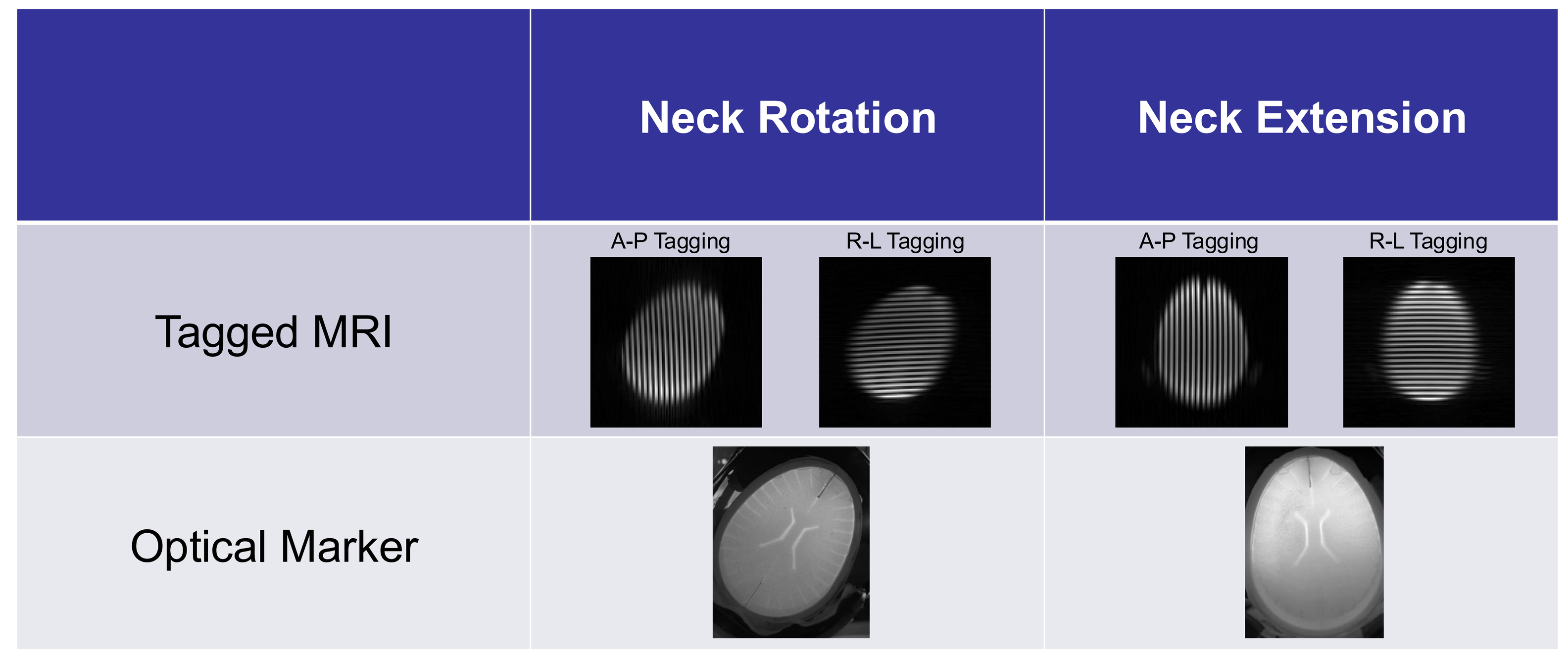

Figure 2. Tagged MRI and optical marker images acquired during neck rotation and neck extension. A–P: anterior–posterior; R–L: right–left.

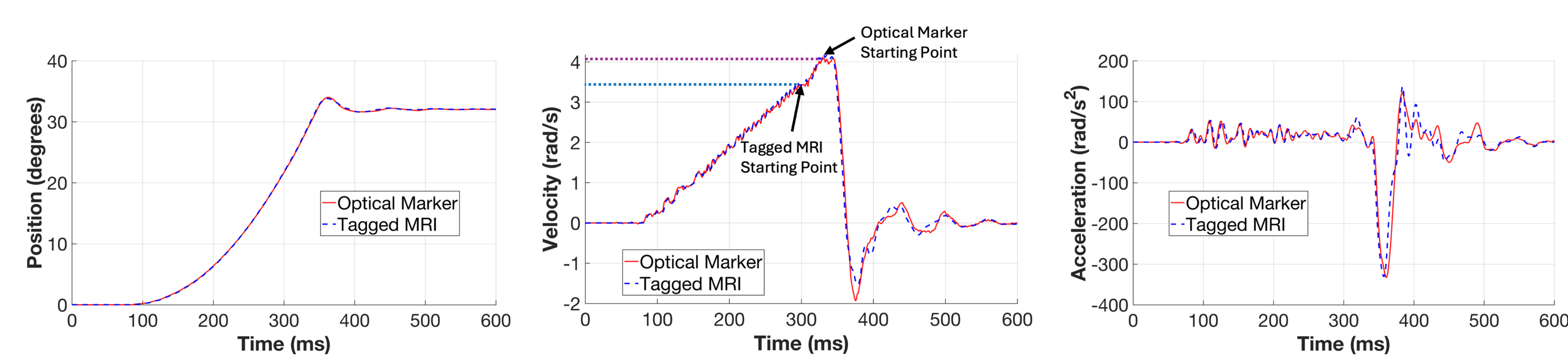

Figure 3. Time histories of angular position, velocity, and acceleration from the optical marker experiment (representative trial) and tagged MRI experiments (mean across trials). Peak angular acceleration was 332.6 rad/s² for the optical marker experiment and 331.4 ± 3.2 rad/s² for the tagged MRI experiments.

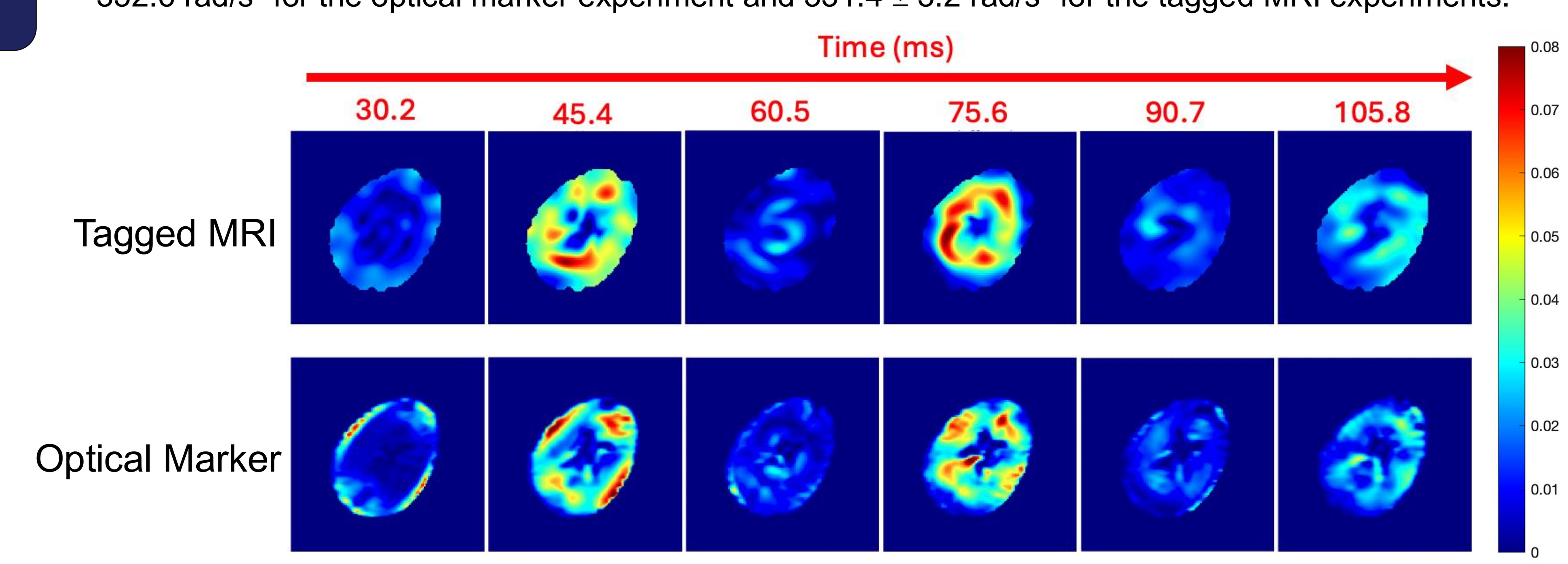

Figure 4. Maximum principal strain (MPS) maps at multiple time points during impact.

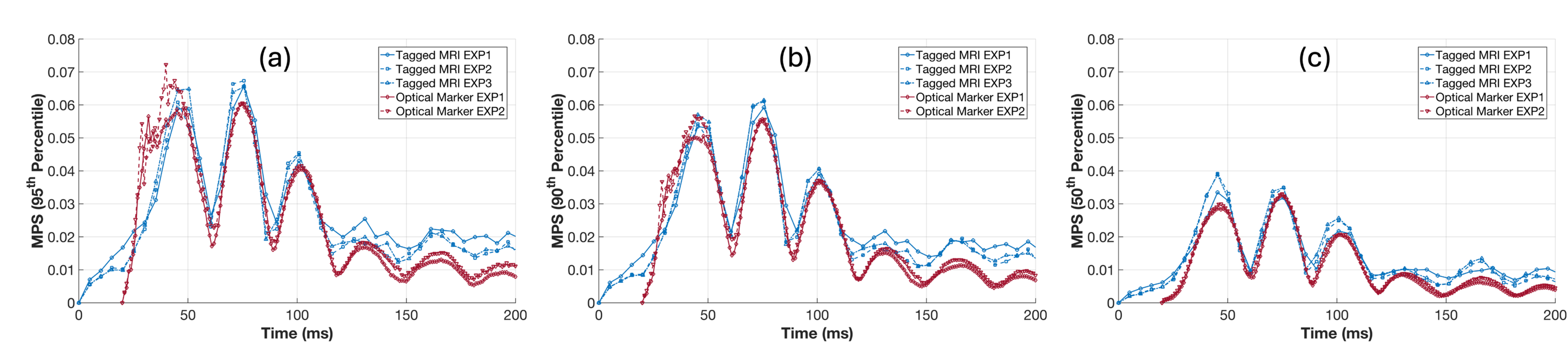

Figure 5. Time histories of maximum principal strain (MPS) during neck rotation experiments: (a) 95th percentile, (b) 90th percentile, and (c) 50th percentile. EXP: experiment.

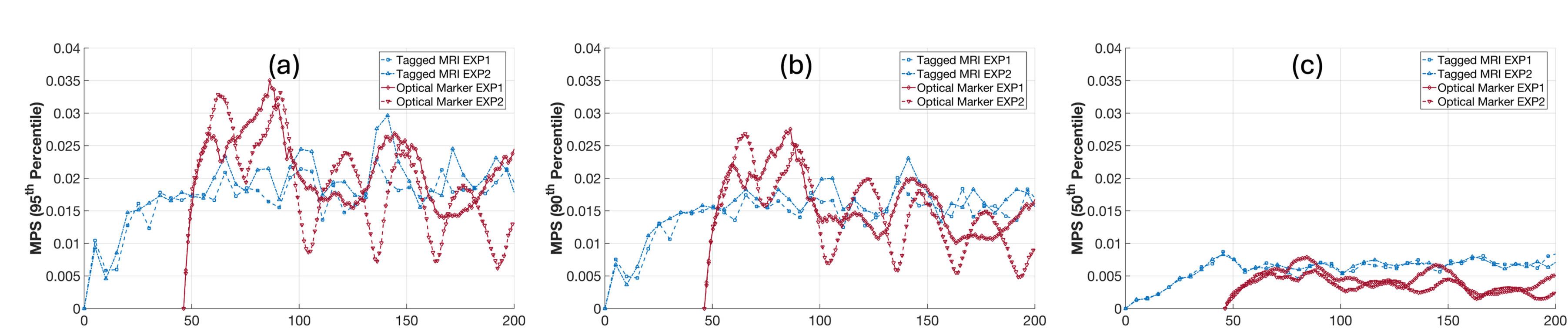

Figure 6. Time histories of maximum principal strain (MPS) during neck extension experiments: (a) 95th percentile, (b) 90th percentile, and (c) 50th percentile. EXP: experiment.

RESULTS

- In tMRI, the phantom exhibited mean peak angular accelerations of 331.4±3.2 rad/s² (rotation) and 255.5±4.7 rad/s² (extension).
- Mean peak 95th percentile MPS values were determined to be 0.070 (rotation) and 0.023 (extension), which were within the range of adjusted *in vivo* MPS values [3].
- Peak strain in both tMRI and optical tracking occurred within 70 ms and dissipated by 150 ms, exhibiting similar patterns to *in vivo* studies.

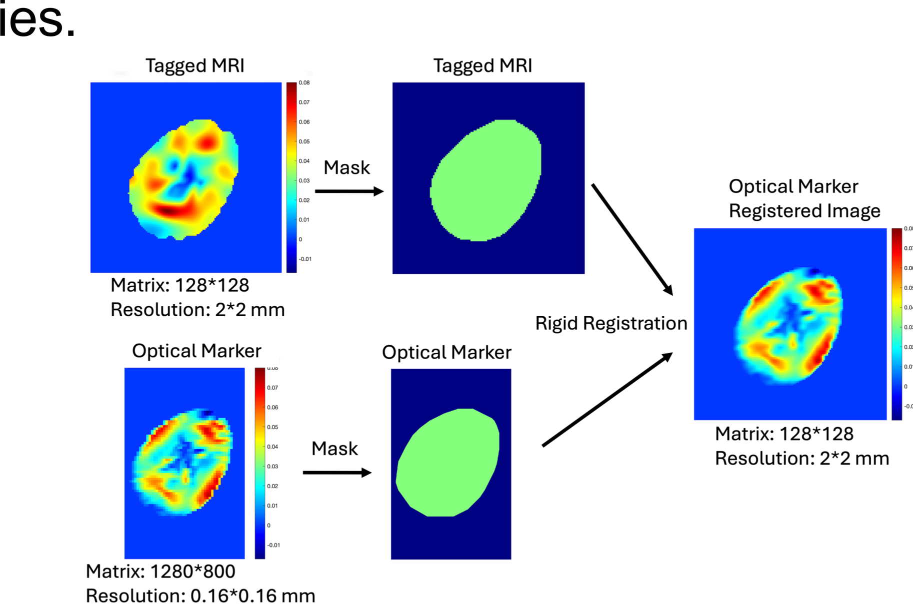

Figure 7. Image registration pipeline for aligning maximum principal strain (MPS) maps derived from tagged MRI and optical marker images.

Tagged MRI	Optical Marker	Pearson Correlation	Spearman Correlation	RMSE	Normalized RMSE
EXP1	EXP1	0.223	0.227	0.019	0.565
EXP1	EXP2	0.322	0.306	0.019	0.555
EXP2	EXP1	0.356	0.375	0.019	0.500
EXP2	EXP2	0.494	0.483	0.018	0.472
EXP3	EXP1	0.310	0.361	0.020	0.538
EXP3	EXP2	0.421	0.469	0.020	0.513

Table 1. Evaluation metrics for aligned maximum principal strain (MPS) maps derived from tagged MRI and optical marker images at the first peak MPS. RMSE: root mean square error.

DISCUSSION

- This study demonstrates the use of a biofidelic brain phantom, combined with tMRI and optical marker tracking, to quantify brain phantom deformation under mild impulsive loading and advance the study of TBI biomechanics.
- Optical tracking offers superior spatial and temporal resolution but is not applicable *in vivo*, whereas tMRI enables *in vivo* studies; together, they bridge phantom and human research to advance understanding of TBI.

REFERENCES

1. Kerwin et al., Front Neurol. 2022
2. Knutsen et al., J Mech Behav Biomed Mater. 2023
3. Knutsen et al., Brain Multiphys. 2020

ACKNOWLEDGEMENTS

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