Motion analysis for shortening scan times by use of affine transformations and HARP in cardiac MRI

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Background

In high resolution cardiac MR imaging, motion artifacts from the intrinsic cardiac motion is traditionally minimized by scanning only during a 60-100 ms time window in mid-diastole (1) as the motion is small during this time period. If the motion is known and is an affine (linear) transformation, it is theoretically possible to scan using a larger time window with a prospective or retrospective motion correction. Thus, it is of interest to examine the extension of the time window where the cardiac motion is affine. The aim of this study was 1) to use a 2D scanning to obtain motion information on a per-pixel basis and subsequently fit the data to an affine transformation, 2) to examine how much of the cardiac motion is affine, and 3) to examine the duration of the time window where the cardiac in a 2D long-axis slice can be described by an affine transformation.

Materials and methods

Seven healthy volunteers were imaged in a Philips Intera 1.5T MR-scanner using a 17 cm surface coil. A long-axis slice was defined along the LAD as identified in 3 short-axis slices from the base middle and apex, respectively.

The sequence used for scanning the long-axis slice was a breath hold, EPI-sequence with an EPI factor of 9 and a reduction of k-space sampling to 35%. CSPAMM tagging distance of 8 mm was used. The scan covered the cardiac cycle starting from 20 ms after the R-wave up to 80% of the cardiac cycle. The frame duration was 35 ms and the total scan duration was approx. 18-20 s (depending on the heart rate of the volunteer).

The motion quantification was based on HARP imaging (2); with HARP imaging, the motion of each pixel from two different time frames can be found. We defined the time window as a period starting at a variable time before mid-diastole and ending in mid-diastole (fig. 1). We defined a region of interest (ROI) as the pixels within the cardiac muscle mass in the long-axis slice. The motion of all pixels within the ROI between the beginning and end of the time window was then calculated. Subsequently, the motion information was fitted to an affine transformation in a least-square sense using singular value decomposition. We defined three different motion correction models: 1) no motion correction, 2) correction for translation, and 3) correction for an affine transformation. We defined the model error as the average length of the motion vector of the remaining motion after correction with a given motion model:

$$e = \frac{1}{N}\sum_{i=1}^{N}\left\|\mathbf{m}_i(\mathbf{x}) - \mathbf{m}_{calc,i}(\mathbf{x})\right\|$$

where e is the model error, N is the number of motions in this time window, **m** is the motion vector and $\mathbf{m}_{calc}$ is the motion vector calculated from the motion model. The model error is a function of the time window used to calculate the motion. A large model error indicates that an image taken with the given motion correction model and time window will have severe motion artifacts.

The maximum model error with acceptable image quality was chosen as 1 mm as determined by visual inspection of images with different synthetic model errors. Subsequently, the maximum time window was identified where the model error was below 1 mm for each of the motion models.

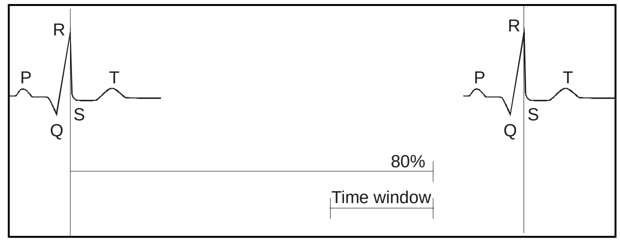

Fig. 1: The location of the time window. The start time is variable and the end time is 80% trough cardiac cycle.

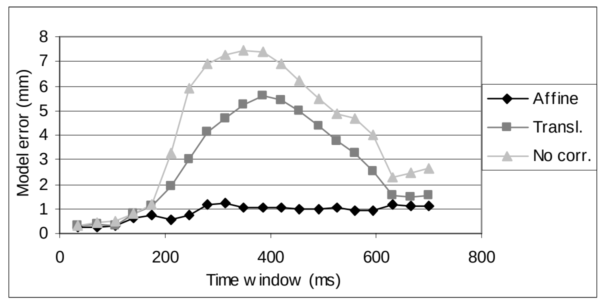

Fig. 2: The model errors for the three motion models as a function of the time window for one volunteer. Notice the low model error with short time windows (i.e. time windows in diastole). Also note the low model error throughout the cardiac cycle for the affine motion model.

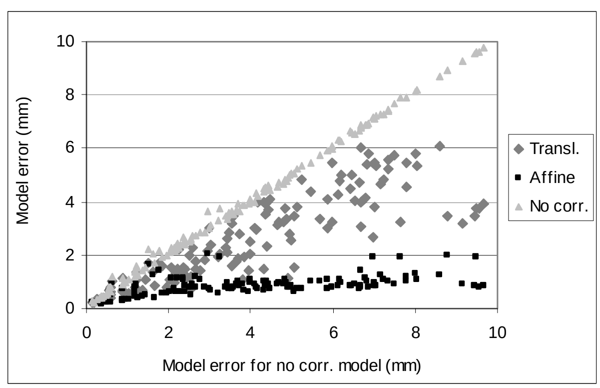

Fig. 3: The model error for all motion models as a function of the model error for the no correction model. It is seen that translation describes some of the motion whereas the affine motion model describes the cardiac motion well.

Results

An example of the model error as a function of the time window is seen for a volunteer in fig. 2. The model error as a function of the model error with no motion correction is shown in fig. 3. The differences between the different model errors were estimated by linear regression. The slope of the difference between no correction and translation was 0.39 meaning that 39% of the model error was removed when correcting for translation. The slope of the difference between no correction and affine was 0.83. For both situations, the probability of a zero slope was less than 0.01.

The time windows where the motion error was below 1 mm can be seen in table 1. The time window of no correction is 100 ms which is in agreement with the standard time window today. It is also seen that correction for an affine transformation expands the time window by a factor 4.5.

	Time window No corr.(ms)	Time window Transl.(ms)	Time window Affine (ms)
Mean	100	140	450
Std.dev.	102	114	309

Table 1: Maximum time windows where the model error is less than 1 mm.

Conclusion

In the present study, we have introduced a new method for analyzing cardiac motion using HARP imaging and affine transformations. We have shown that most of the cardiac motion in a long-axis slice is affine and that the cardiac motion in this slice may be described by a simple affine transformation in a time window substantially larger than the time window used presently in high resolution cardiac imaging. The presented techniques have potentials for detecting the motion of cardiac structures and the data obtained indicates that either prospective or retrospective correction of the affine motion would allow a substantial increase of the time window for high-resolution cardiac imaging with subsequent decrease of the scan time.

References

1. Botnar RM et al. Circulation, 99(24):3139-48, 1999
2. Osman NF et al, Magn Reson Med, 42(6):1048-60, 1999