**Supplementary Methods**

**Image acquisition and preprocessing**

All image data were acquired with a Siemens 3T Trio scanner (Siemens, Erlangen, Germany). All participants were asked to abstain from smoking, caffeine and other stimulants from midnight the night before his/her scan. A high-resolution 3D T1-weighted image was collected using a magnetization-prepared rapid gradient echo (MPRAGE) sequence (TR/TE= 1670/1.89 ms, FA = 9°, voxel size = 1×0.98×0.98 mm3, 208 sagittal slices) for anatomical reference. RS-fMRI data collection was performed for a 6 min and 53 s using an echo planar imaging sequence (TR/TE = 3500/30 ms, flip angle (FA) = 90°, voxel size = 1.9×1.9×3.5 mm3, 35 axial slices). During RS-fMRI scan, participants were instructed to keep their eyes closed and to not think about anything. To ensure that participants did not fall asleep, they were reminded to stay awake through microphone immediate before the RS-fMRI acquisition. After scanning, a simple questionnaire was administered to confirm they had not fallen asleep.

Image preprocessing was performed using SPM12 (http://www.fil.ion.ucl.ac.uk/spm) and DPARSFA toolbox (http://rfmri.org/DPARSF).(Chao-Gan & Yu-Feng 2010) After discarding the first four volumes for each subject, images were corrected for slice timing and head motion. To reduce motion-related artifacts, the data were checked with the following criterion for excessive head motion; (i) six motion parameters < 1.5 mm or 1.5° in any direction, (ii) mean framewise displacement (FD) and the root mean square (RMS) of motion proposed by Power et al.(Power *et al.* 2012) and Van Dijk et al,(Van Dijk *et al.* 2012) respectively < 2 standard deviations from the group mean (in our case, <.31 mm and <.083 mm, respectively),(Yan *et al.* 2013) and (iii) mean FD proposed by Jenkinson et al.(Jenkinson *et al.* 2002) <.25 mm.(Parkes *et al.* 2018) The remaining data were co-registered to the T1 structural image of each individual subject and then the coregistered images were segmented into gray matter, white matter, and cerebrospinal fluid (CSF). Next, the images were further regressed out of the following nuisance variables: six motion parameters and their first derivatives, head motion scrubbing regressors (FD >0.5, one volume before and two volumes after the bad time point as the default option), five principal components estimated from both white matter and CSF regions using an anatomical component-based noise correction (aCompCor) method,(Behzadi *et al.* 2007) and two polynomial trending terms for linear and quadratic trends. The residual images were normalized to the Montreal Neurological Institute (MNI) space and were then smoothed with a 6 mm full-width half-maximum Gaussian kernel. Finally, time series were band-pass filtered (0.01–0.1Hz) to reduce the effect of low frequency drift and high frequency physiological noise. Like pervious MDMR studies,(Shehzad *et al.* 2014; Satterthwaite *et al.* 2015; Sharma *et al.* 2017) the preprocessed data were finally down-sampled to 4-mm isotropic voxels to allow for computational feasibility.

**Reference**

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**Supplementary Figure 1.** MDMR-based CWAS analysis flowchart. For each gray matter voxel and each participant, a connectivity map was generated by temporal correlations between a given voxel and every other gray matter voxel. Next, the distance between every pair of participants’ connectivity maps was calculated. Then, MDMR was applied to this distance matrix to evaluate the multivariate pattern of connectivity associated with the discount rate across participants, while adjustments were made for age, sex, and head motion. This analysis produced a pseudo-F statistic, and an accompanying p-value was obtained by 5000 permutations. This procedure was repeated for each gray matter voxel to yield a whole-brain voxel-level significant map, which was ultimately thresholded at a height threshold of z >3.1 and a corrected cluster probability of p <0.05.